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ADVISORY GROUP FOR AEROSPACE RESEARCH & DEVELOPMENT

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Neck Injury in Advanced Military Aircraft Environments

NORTH ATLANTIC TREATY ORGANIZATION

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ADVISORY GROUP FOR AEROSPACE RESEARCH AND DEVELOPMENT

(ORGANISATION DU TRAITE DE L'ATLANTIQUE NORD)

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NECK INJURY IN ADVANCED MILITARY AIRCRAFT ENVIRONMENTS

Papers presented at the Aerospace Medical Panel Symposium held in Munich, Germany,
from 24 to 28 April 1989.

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PREFACE

Neck injury from excessive force may become more common with the extension of high-acceleration performance envelopes and improved escape system operation in training and combat aircraft. The introduction of helmet-mounted equipment may further increase the risk of injury from these sources as well as from accidents in both fixed and rotary wing aircraft.

This Symposium discussed the extent of this risk and its control through the design of helmet-mounted devices, protective systems and aircrew training and conditioning.

* * *

PREFACE

Les lésions du cou résultant de l'application de forces excessives risquent de devenir de plus en plus courantes avec l'accroissement des performances en accélération dans le domaine de vol et les améliorations qui seront apportées au fonctionnement des systèmes d'évacuation des avions d'entraînement et de combat. L'adoption de matériel monté sur le casque pourrait aggraver le risque de lésions ayant les mêmes origines ainsi que le risque d'accidents survenant aux aéronefs à voilure fixe et à voilure tournante.

Ce Symposium a examiné l'importance de ce risque ainsi que les moyens qui sont à mettre en oeuvre pour le maîtriser par le biais de la conception des dispositifs montés sur le casque, les systèmes de protection, et la formation et la préparation psychologique du personnel navigant.

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TECHNICAL EVALUATION
AEROSPACE MEDICAL PANEL MEETING
"NECK INJURY IN ADVANCED MILITARY AIRCRAFT ENVIRONMENTS"

D J Anton
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Now in building of chaises I tell you what,
There is always *somewhere* a weakest spot,
In hub, tire, felloe, in spring or thill,
In panel, or crossbar, or floor, or sill,
In screw, bolt or thoroughbrace,- lurking still
Find it somewhere you must and will,
Above or below, or within or without,-
And that's the reason, beyond a doubt,
That a chaise *breaks down*, but doesn't *wear out*.

From 'The Deacon's Masterpiece or
The Wonderful 'One Hoss Shay'
Oliver Wendell Holmes (1809-1894)

The 67th Aerospace Medical Panel Meeting and Symposia were held in Munich between 24th and 28th April 1989. Two days of the meeting were devoted to the topic of neck injury in advanced military aircraft environments. The theme of the meeting could be summed up as "Is the neck the weak link in the high G environment". Two sessions were held: one on epidemiology and pathology, the second on the dynamics of head and neck motion.

Epidemiology and Pathology

Like the ordinary chaise referred to in Oliver Wendell Holmes's poem, the human neck is susceptible, if appropriately stressed, to breaking down. This breakdown can take the form of either vertebral fracture or, more commonly, soft tissue injury. Unlike the chaise, there is also some evidence to suggest that the bearing surfaces of the neck might wear out as a result of repeated stressing. The first session of the meeting, on epidemiology and pathology, contained five papers that reviewed the evidence for the occurrence of both acute and chronic neck injury in high performance aircraft. Two papers were devoted to radiological investigation of the cervical spine and data analysis in cervical trauma and a third paper reviewed electronystagmographic findings following cervical injury.

The data presented showed that a significant problem existed with neck pain and injury in high performance aircraft as compared with other combat aircraft. Authors reported incidences of cervical injury between 30-50% of the exposed aircrew population. The more agile the aircraft, the greater the number of aircrew reporting symptoms. There was conflicting evidence on the effect of age on the incidence of injury. Injury appeared to be more prevalent in the training environment, although this statement may reflect the effects of inexperience, and the greater than normal exposure of the instructors to high G manoeuvering.

Several authors remarked on the importance of neck muscle conditioning as a means of preventing injury. The importance of both formal physical training programs, and neck muscle 'warm up' prior to flight was stressed, although no data was presented analysing the value of different training regimes.

A matter of considerable concern was the possibility that repeated episodes of trauma could lead to chronic degenerative changes in the cervical spine. The Belgian Air Force Medical Service presented details of a cervical spinal screening program where all pilot candidates are X-rayed *ab initio* and at intervals of five years. The aim is to compare the results of F16 pilots with a control group of pilots not flying the F16. Such a survey is vital if the question of whether high-G flying provokes cervical spondylosis is to be answered. Similar surveys need to be done in a comparable manner in other Air Forces in order to obtain a larger database and raise the level of confidence in the survey outcome.

Dynamics of head and neck motion

The second session was devoted to twelve papers on aspects of the dynamics of head and neck motion and the use of computer models in head and neck motion simulation. Two particular topics were covered: the response of the head and neck to flight acceleration and impact loads, with and without the effects of added mass, and the assessment of cervical injury risk using both computer simulation and biomechanical dummies.

A number of papers highlighted the considerable effort that is being made to define and model the acceptable mass and mass distribution characteristics of aircrew helmets and helmet mounted equipment. Some of this work is based on computer simulations of neck response that are validated against live subject experiments in the range of voluntary tolerance. This work is always difficult to extrapolate into the range where injury is expected and cadavers have been used to expand the experimental range.

The results of computer simulation appear to be rather conservative in the prediction of cervical injury risk in impact environments. There is still a need for further careful and thorough studies of the cervical spine in accidental impacts!

The paper by Tarriere et al reviewed the occurrence of cervical fracture in road traffic accidents and in a series of cadaver experiments, and concluded that such fractures were extremely uncommon, particularly in the absence of head injury. This observation echoes the comment of Goldsmith and Ommaya (1984) that the cervical region is less frequently traumatised in impacts than the head. Of topical interest was Tarriere and his colleagues observation that women ran a two to threefold greater risk of neck injury than men. This finding was not elaborated on but may reflect the lesser muscular development and smaller vertebral size generally seen in women.

Conclusion

This was a useful meeting that that served to highlight the need for two particular areas of research:

1) Further studies on the epidemiology of acute and chronic neck injury arising from in-flight manoeuvering loads.

2) The continuing need for improved understanding of the mechanisms of neck injury on impact.

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NECK INJURY IN ADVANCED MILITARY AIRCRAFT ENVIRONMENTS

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Introduction

The advent of high performance aircraft has produced a series of challenges in the field of aircrew protection. Amongst these has been a growing interest in the effects on the neck of additional mass on the head; and increased effort has been directed in recent years towards the development of light weight helmets to minimise fatigue and maximise head mobility under higher levels of aircraft acceleration. This trend to lighter weight helmets was welcomed by those charged with the evaluation of escape systems, as concern had been voiced that the new generations of ejection seat might cause more neck injury, due to the higher forces imposed on the seat occupant by reducing the time to full parachute inflation. Lighter weight helmets would theoretically help to reduce the injury risk.

The concern about the problem of neck injury on ejection was partially addressed by AGARD Working Group 11, but their 1984 report concluded that: 'there was little reliable evidence to suggest that head and neck injury was a significant problem on 'within envelope' ejections on modern ejection seats'.

As a consequence, however, of the background work to the WG 11 report, instances were noted of cervical fracture occurring in flight; normally involving the unaware crewmember, and generally following the application of high manoeuvering loads as part of air combat, or to avoid birds or terrain. These observations have been made in the United States Air Force, the Royal Norwegian Air Force, and the Royal Air Force, suggesting that the problem is more common than had previously been realised.

With the deployment of aircraft capable of sustained high-G, complaints about acutely stiff and painful necks also became more common. Anxiety was expressed by clinicians involved with the treatment of aircrew, that repetitive vertebral trauma might lead to an increased premature appearance of cervical osteophytes, and that in turn might lead to the affected aircrewman being withdrawn prematurely from fast jet flying. As a corollary to this, arguments were raised for introducing spinal screening of aircrew candidates in those countries where such a procedure was not already in place.

The introduction of night vision goggles in helicopters and subsequently in fast jets, and the postulated introduction of helmet mounted sighting and display systems, threatened to reverse the trend towards lighter helmets and exacerbate the injury risk. Discussion took place about the balance between operational need and safety, and it became apparent that few firm or useful data existed to indicate either maximum allowable added mass on the head, or what were the acceptable head mounted mass distribution characteristics.

Some help in analysing the mass implications was available from computer modelling, but many of these models suffered from the limitation that they tended to deal only with shear and torque forces at the cranio-cervical junction, and with a limited number of effects of pure antero-posterior bending (flexion and extension). What epidemiological data were available suggested that shear and torque injuries at C1/2 were relatively uncommon, but frequently fatal: simple compression fractures at C5 being more the norm. Furthermore, the models were only claimed to be predictive of structural rather than soft tissue damage, although the soft tissue injury might be of greater operational consequence, in that it might limit the number of aircrew fit to fly at any one time. It had become obvious that there was a need for considerable further information, and also a need to develop some form of classification with a view to more precise definition of research requirements. There was a further urgent need for epidemiological and accident reconstruction data to validate the computer modelling.

Classification

The following is offered as a working classification of neck injuries in aircrew; it is by no means comprehensive, and it is arguable that complete or partial ligamentous disruption should be included with fracture, rather than being included under the heading of soft tissue injury.

1. Acute

   a) soft tissue neck injury, including ligamentous disruption, arising in-flight.

   b) vertebral fracture arising in-flight.

   c) soft tissue injury arising from acute loading in crash or assisted escape environments.

f) fractures arising from acute loading in crash or assisted escape environments.

2. Chronic

a) chronic degenerative disease (cervical osteophytosis) arising from exposure to aircraft manoeuvering loads. (? existence).

b) chronic degenerative disease (cervical osteophytosis) arising from damage sustained in aircraft ejection or crash.

1a. Soft tissue neck injury arising in-flight.

Some information is now available on the incidence of soft tissue neck injury in aircrew populations. Aghina (1984) has reported that F16 pilots had eight times more neck symptoms than F104 pilots, and that during an observation period of six months, [illegible] of the F16 pilots had experienced neck symptoms. Since these reports relate to a population flying with relatively light weight helmets, it is clear that increasing helmet mass is likely to increase both the intensity and frequency of neck symptoms. The problem of neck injury in the F16 may be exacerbated by the sitting position, which because of the ejection seat requires a compensatory flexion of the cervical spine.

Vanderbeek (1988) reported on a sample of 437 pilots flying three aircraft types and showed that increased aircraft performance was associated with increased neck injury, and that increased age was associated with increased prevalence of major neck injury.

The significance of soft tissue neck injury occurring as a result of high-G flying is that it can cause an important loss of aircrew effort to the flying task, and in a way that is not readily susceptible of medical treatment. It thus becomes a condition that could have a significant effect on the ability to generate fighter sorties in time of conflict.

Knudson et al (1988) make precisely this point in their survey of high +Gz induced neck pain and injury in aviators from the USN Pacific Light Attack Wing. They found the [illegible] of F/A-18 aviators surveyed reported neck pain with high +Gz. Out of [illegible] pilots reporting neck injury, 11 required removal from flight status for an average 3 days. The inability to function effectively during high G flight, and the impact of lost pilot days highlighted the need for further studies into the prevalence and solutions for +Gz induced neck injury.

Some authors recommend that aircrew should conduct neck strengthening exercises as a routine, and also engage in neck 'warm up' exercises immediately prior to the start of high-G flying. This advice has apparently been adopted in some Air Forces although data validating the usefulness of neck strengthening and neck 'warm up' techniques are lacking. The quoted papers are useful starting points in the unravelling of the incidence of soft tissue neck injury. More information is needed, however, in the form of larger scale, controlled studies, using an agreed protocol and soft tissue injury classification system. In-flight research to document what it is exactly that pilots do with their heads in combat is also required, preferably allied with studies to define the influences that ejection seat, helmet and personal equipment design have upon head mobility.

1b. Neck fracture arising in-flight.

This subject is of considerable interest since the fractures that have been reported have occurred at very much lower acceleration levels than are conventionally assumed to be required to provoke injury.

Schall (1989), and Andersen (1988), have reported on cervical fractures sustained whilst air combat manoeuvering. The case Andersen reports is typical of the circumstances of injury. A flight surgeon flying a mixed exercise, including an interception, from the back seat of an F16B, hands over the control of the aircraft to the front seater. Following this, the flight surgeon relaxes in his seat and turns his head maximally over his left shoulder to look for the opponent. As he does this the front seater abruptly institutes an 8G climbing turn. The flight surgeon is caught completely unawares, momentarily losing consciousness, and his next recollection is of being jackknifed in the cockpit under sustained high G. The flight is continued, but in the ensuing hours after landing nausea and neck pain developed. Equivocal radiographic evidence of a C6 compression fracture was obtained, together with evidence of C5/6 ligamentous injury as shown by disc space widening and minor posterior slippage of C6 with respect to C5.

It is of no surprise that if injuries like this can occur as a result of in-flight loads, they do not occur more commonly as a result of emergency escape. Clearly, both the peak G and the onset rate for Andersen's case do not significantly exceed 8G, or an onset rate of 6-10G sec-1, figures that are comfortably below those associated with escape system usage.

Schmorl (1971) considered the question of vertebral fracture following minor trauma, noting that at times quite minor trauma may cause collapse of a healthy vertebra. He was of the view that the uncoordinated contraction of various groups of back muscles with resulting paradoxic fixation and rotation of the spine, appeared to be the

mechanism responsible. Schmorl also noted that pain and other neurologic symptoms appeared some time after the trauma, perhaps because of the detachment of an impacted fragment, or the gradual development of a haematoma. Thus, acute loading under high G, at onset rates that approximate times for reflex contraction, coupled with an off axis position of the neck, may be an explanation for these injuries. If Schmorl's explanation is correct, it would suggest that neck fracture occurring as a result of inflight loads is a phenomenon that is relatively independent of helmet mass.

The question of why such injuries do not occur on emergency escape remains unanswered, although the more rapid onset of acceleration, together with the head being directed forward at the time of seat initiation may act as a form of protection.

1c. Soft tissue injury arising from acute loading in crash or assisted escape environments.

A degree of soft tissue injury as evidenced by a 'stiff' neck post ejection, or crash, is not uncommon. Indeed at the minor level it is probably so common that it is frequently not properly documented. Royal Air Force ejection experience suggests that there is a minor neck injury rate of approximately 30%, although this figure has not been validated. It is important that more attention should be devoted to soft tissue injury arising from ejection, as emergency escape occurring within the safe escape envelope exposes the aircrewman to a broadly known force environment. Ejection testing provides statistical information that enable the bounds of the acceleration environment to be set, particularly for the straight and level case. Knowledge of the ejection test data, the circumstances of the actual ejections, the characteristics of the head mounted equipment and the injury outcome can thus constitute an important epidemiological data base from which general statements of injury risk can be made.

A few 'within envelope' ejections are fatal, and these pose a challenge to both the investigator and the pathologist. It is extremely important that particular attention is paid to the possibility of neck injury, even where the evidence is that neck injury is not the proximate cause of death. Dissection of the neck is time consuming and probably cannot be justified as a routine autopsy feature, but in those cases of unexplained fatality on 'within envelope' ejections it is essential that evidence of neck injury be sought if a complete investigation of the accident is to be made. The following is a typical example of the sort of case that occurs:

Case No 185. (Anton 1984). The aircraft was one of a pairs formation. They had been briefed to do a loose article check at some stage of the flight. At 400kts in company with the first aircraft the subject aircraft rolled a full 360 degrees and then rolled again. In the second roll it pitched nose down. The canopy was seen to detach and a flash, possibly due to the rocket motor of the ejection seat, was seen. The instructor from the rear seat of the subject aircraft was recovered drowned beneath an apparently normally deployed parachute. He had not accomplished any post ejection drills. Investigation showed that the aircraft was yawing markedly at the time ejection was initiated, which was probably 'within envelope'. Investigation also revealed evidence of lateral extraction from the ejection seat. At autopsy the cervical spine and cord were dissected out, but there was no evidence of injury to the brain, or spinal column or cord, although there was bruising in the right paravertebral muscles. Other autopsy evidence indicated that death was due to drowning. The observation that the pilot had failed to complete any of his post ejection survival drills gave rise to the view that he may have been incapacitated as a result of the ejection forces. In this case, however, the care taken to dissect the neck failed to produce positive evidence of cord damage, although the haemorrhage in the paravertebral muscles showed that the neck had been subject to trauma. This case highlights a further difficulty in that drowning, as an asphyxial death, produces congestive changes in the brain that preclude determination of the effects of all but the grossest results of inertial injury.

1d. Fractures arising from acute loading in crash or assisted escape environments.

This topic was specifically addressed by AGARD Working Group 11. The evidence then available revealed a wide variation in the reported incidence of head neck injury on ejection. Much of this variation was due to differences in criteria for inclusion of injury between different Air Forces, not all distinguishing in their reporting between 'within envelope' ejections and those outside safe escape capability. Both the USN and the RAF reported incidences of severe or osseous neck injury of between 1-2% on 'within envelope' escape, the injuries generally being simple compression fractures. Much less has been written in the open literature about the types and frequency of fractures occurring in crashes, particularly of helicopters. This area is of considerable interest since it might be expected that the addition of added mass for night vision or display systems would change the frequency or types of cervical fractures seen in fatal accidents.

2. Chronic Injury

Schmorl (op cit) also dealt with the relationship between trauma and what he termed spondylosis deformans (osteophytosis). He posed three questions which have still only been partially answered, particularly for the cervical spine.

a) Does traumatic spondylosis deformans exist without vertebral fracture?

b) Does traumatic spondylosis deformans occur in connection with a vertebral fracture?

c) Can an existing generalised spondylosis deformans be exacerbated by trauma?

The concept of chronic degenerative disease of the cervical spine arising as a consequence of either single acts of major trauma or repeated instances of minor insult is common. Schmorl emphasised abnormalities in the peripheral fibres of the annulus fibrosus as the initiating factor. He suggested that breakdown occurred at the site of the outer annular fibres attachment to the vertebral rim. This allowed displacement of disc material producing in turn, displacement of the overlying longitudinal ligament, and stress at the site of the ligaments attachment to the vertebra. Osteophytes may then develop at these stressed areas. Lipson and Muir (1980) looked at the effect of ventral nuclear herniation in rabbits. This showed that the technique of ventral nuclear herniation of discs in the lumbar spine could reliably produce osteophytes which arose from proliferating inner annular fibres. These fibres underwent metaplasia into cartilage, calcified, and then changed to osteophytes through an endochondral ossification sequence. All of the experimental work that has been conducted appears, however, to relate only to the lumbar spine and it is not clear to what extent the findings can be extrapolated to the neck.

Schmorl raised the question of a generalised spondylosis deformans being exacerbated, or even caused by trauma, only to dismiss it. This appears reasonable as all the available evidence points to the disease arising as a result of local changes. It is true that disease is seen at more than one level in some, generally more elderly individuals, but this is not per se an argument in favour of a generalised disease occurring as a result of trauma.

Degenerative changes of the intervertebral discs of the cervical spine are also common, particularly after the age of 40, and affect 70% of patients over the age of 70. (De Palma et al 1970). The most commonly involved site is the intervertebral disc at C5/6, followed by C6/7, these being the points of maximal flexion. In contrast, the intervertebral disc at the C2/3 level is least often affected. Frequently, associated changes occur in the Joints of Luschka and osteoarthritis of the apophyseal articulations is also more common in the middle and lower cervical spine.

2a. Chronic degenerative disease (cervical spondylosis) arising from exposure to aircraft manoeuvering loads.

2b. Chronic degenerative disease (cervical spondylosis) arising from damage sustained in aircraft ejection.

MacKenzie Crooks (1970) provided figures that showed that the incidence of radiologically diagnosed cervical spondylosis (spondylosis deformans) is higher in RAF pilots than in the civil population. Subsequently, Anton and Cave (1986) conducted an analysis of the immediate post ejection cervical Xrays of 68 aircrew (postulating that ejection acted as a randomisation process for selecting an aircrew population). This study used the following classification for the analysis of the Xray changes:

0. No evidence of change (normal).

1a. Minimal loss of lordosis and/or minimal disc space narrowing at C5/6.

1b. Small osteophytes posteriorly on the lateral lip of uncovertebral Joints of Luschka.

2. Moderate. Osteophytes at several levels. Marked disc space narrowing. Obvious osteophyte encroachment on neural foraminae.

3. Severe. Two or more levels involved by large osteophytes and marked disc space narrowing.

Fifty of the 68 subjects exhibited no evidence of change, 15 showed grade 1a changes and 3 showed grade 1b change. The mean age of the group was 32 years which is similar to the 30 yrs +/-6.5 yrs, age of the 225 aircrew surveyed in the RAF survey of ejections 1968-1983 (Anton 1986), and the mean age of 2000 RAF aircrew in the 2000 aircrew survey, 30.8 yrs +/-6.85 yrs (Bolton et al 1973). Far fewer Xray changes are noted in this later group than in MacKenzie Crooks original observations, which suggests a need both for repeat observations and well matched controls. If the Anton and Cave figures are representative, a larger sample than is available from RAF fast jet aircrew would be required, and this helps to make a case for joint studies with other Air Forces using agreed protocols.

MacKenzie Crooks also observed that the incidence of radiological changes of cervical spondylosis (70%) was much higher in aircrew who had ejected, than in both a reported series in civilians. (Friedenburg and Miller 1963), and in a control series of pilots who had not ejected. This finding of an increased incidence of change in the group who had

ejected would suggest that trauma to the cervical spine is indeed common on ejection and carries a long term sequel. Whether, however, these radiological changes are clinically significant in aircrew remains to be established, and it would be useful to see a repeat of MacKenzie Crook's study.

If it can be demonstrated that flying high performance aircraft produces an increased incidence of radiologically diagnosed cervical spondylosis, the question of the clinical significance of the finding still has to be addressed. Xray changes frequently occur without clinical symptoms, and except in the case of large posterior osteophytes, which could theoretically fracture under load and compromise the integrity of the spinal cord, the Xray findings should not of themselves be a reason for rejection of the aircrewman from high performance aircraft flying.

Prediction of neck tolerance to trauma.

Tolerance criteria for neck injury are not well established, a situation which contrasts with head injury. Goldsmith and Ommaya (1984) list five reasons for this lack of information:

1) there are fewer investigations of neck injury, since the cervical region is less frequently traumatised than the head,

2) neck injuries do not exhibit the spectrum of severity as is the case with head injury, injuries tending to be either minor or catastrophic,

3) neck response is crucially dependant upon the orientation of the head and neck and also upon the direction of load,

4) no effective scaling relationship from animal data has been developed for the neck.

To the above may be added a fifth reason:

5) the much more complicated structure and response of the neck does not readily lend itself to the development of either mechanical or mathematical analogues.

Volunteer and Cadaver experiments

There is an encyclopaedic data base on the response of the human neck, to predominantly -Gx impacts, within the range of accelerations that are tolerable to volunteers. Much of the credit for this must go to Ewing and Thomas and their colleagues at the United States Navy Biodynamics Laboratory, New Orleans. Unfortunately, however, extrapolation of these data to higher levels of acceleration is limited by the non linear response of the neck and the difficulty in predicting the position of the neck at the start of the impact. Mertz and Patrick (1971) undertook a series of experiments in which human volunteers were subjected to static and dynamic environments which produced non-injurious neck responses for neck extension and flexion. Cadavers were then used to extend the data into the injury region. Analysis of the data from the volunteer and cadaver experiments indicated that the magnitude of the moment about the occipital condyles was the critical injury parameter for both extension and flexion. The authors developed neck response envelopes for the performance of mechanical necks in flexion and extension, and tolerance levels for the neck in flexion and extension. These are shown in Figures 1 & 2.

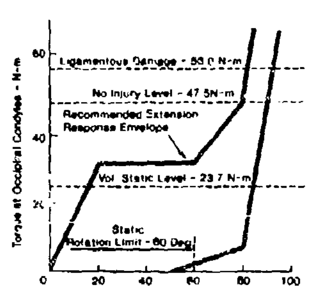

FIG 1 HEAD-NECK RESPONSE ENVELOPE FOR EXTENSION AND VARIOUS TOLERANCE LEVELS
Redrawn from Mertz & Patrick (1971)

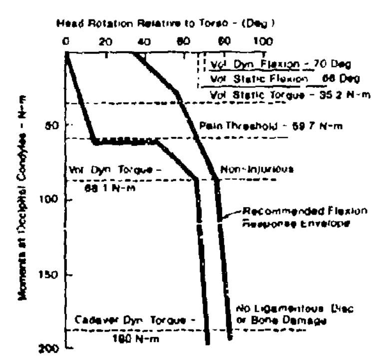

FIG 2 HEAD-NECK RESPONSE ENVELOPE FOR FLEXION AND VARIOUS TOLERANCE LEVELS
Redrawn from Mertz & Patrick (1971)

Some work has been undertaken looking at the safety aspects of the addition of helmets and other masses to the head. Muzzy et al (1986) describe a series of -Gx acceleration exposure experiments using Navy volunteers in which the dynamic response of the head was measured as a function of mass distribution variations. The kinematic response was measured for each subject with no mass addition, a helmet and weight carrying frame, and a helmet with weights positioned on the frame mid sagitally. There was approximately a 30% addition to the head mass with these weights. Computer modelling of the head neck response was used to predict the severity of exposure with mass additions. The results of their investigation validated the models used for predicting mass effects and showed that maximum angular travel was the first safety parameter to approach the established limit threshold. (Two subjects approached this limit at -6Gx with weights). Subsequently, the condyle torque load limit was approached at -8Gx, again with the subjects wearing weights. An important finding of the investigation was the fact that due to response variability between subjects, it was not possible to define discreet figures for safety. The authors suggest that a series of safety parameters have to be adhered to as some of their subjects reached angular displacement and torque limits, whereas others did not reach any limit at all.

It is of note that current UK helmets and oxygen masks add a total mass to the head of approximately 2.0kg, compared with the 1.35Kg added mass involved in the experiments of Muzzy et al!

Computer modelling

Because of the limitations of volunteer experiments, increasing attention has been directed towards the use of computer models such as the head spine model used at the United States Air Force Armstrong Aerospace Medical Research Laboratory (USAF AAMRL). The version currently in use represents the neck by two parallel three dimensional beam elements; one of these having nonlinear viscoelastic axial load deformation characteristics representing the cervical spine. The other element has nonlinear bending behaviour and is used to account for the nonlinear stiffening effects of the soft tissues of the neck. (Privitzer & Belleccerri 1987).

Validation of the head spine model has been pursued at AAMRL for a number of years using both operational ejection spinal injury data and the comparison of model prediction with experimental data. Much of this validation has, however, necessarily been centred around the response of the thoracolumbar spine which is constrained anatomically to a much more limited range of motion than the neck.

It regrettably remains true that although computer modelling provides vital insights into the mechanical behaviour of the neck, it is not yet sufficiently sophisticated to be a method upon which one can rely when considering the acceptability of an actual piece of operational head mounted equipment.

Mechanical dummies

An alternative to the modelling approach is to use sophisticated dummies with 'state of the art' neck instrumentation. One such example is the Hybrid III dummy equipped with upper and lower six degree of freedom neck transducers. These transducers measure forces along the three orthogonal axes and the moments about these axes. Such a tool is clearly valuable as it enables forces to be measured under a variety of impact conditions and with a variety of head loading states. It still, however, is subject to the known deficiencies in dummy neck behaviour and requires extensive validation against both volunteer and cadaver experiments before it is possible to translate impact measurements into meaningful assessments of risk.

Seemann et al (1986) compared the standard Hybrid III neck against human volunteer neck response to -15Gx impacts, and also to +y,+z, and -x,+y impacts using the NBDL data base. They concluded that the Hybrid III is much too stiff to respond in a human like manner to -x and +z, but has a remarkable and unexpected similarity to human neck motion for +y and -x,+y impacts. Modifications to the linkage modelled between the Hybrid III head and neck were tried and resulted in an improvement to the model response. The modelling indicated that a physical relocation of the model/neck torso joint would result in much improved dummy -Gx response.

In a further paper Muzzy et al (1984) compared human versus manikin head and neck response to +Gz acceleration, exposing subjects to peak accelerations ranging from 3G to 12G at onset rates from 100 to 1200G.sec-1. It was observed that the human head response appeared to be very sensitive to initial head orientation and position. Two types of response were observed with the humans; one response which was primarily flexion of the head and neck, and another in which the head exhibited significant extension followed by flexion. The dummy head response was only in flexion. It was concluded that further work was required analysing the effect of the initial position on head neck response.

It is clear from the forgoing that an enormous amount of work remains to be done before acceptable mathematical or mechanical models can be used for accurate prediction of human head neck response to impact; with or without the addition of added mass. Unfortunately, the specifications for current helmet mounted systems will have to be completed long before much of this work is undertaken, so there is an urgent requirement for the intelligent guess as to what is acceptable!

Conclusion

This overview of the scope of the problem of neck injury in high performance aircraft aircrew underlines the large gaps in our clinical and biomechanical knowledge. It also highlights the difficulties involved at present in giving the engineers of helmet mounted systems unambiguous guidelines as to maximum mass and mass distribution characteristics. The advent of sophisticated head mounted systems offers a considerable challenge in both aeromedical research and equipment design.

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Trends in Helmet Mounted Equipment

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ABSTRACT

The paper gives an overview of the development of the flying helmet from being merely a method of supporting a microphone and headset to an integrated part of an advanced aircraft's avionics and weapon system. The operational needs of the modern fast-jet aircraft are discussed and it is reasoned that the pilot must be equipped with an 'electric hat' of some complexity to be mission effective. Electronic and optical technology is being exploited to the maximum to give the pilot the data he requires, when and where he needs it most, yet still allow him to survive physically his cockpit environment to fly and fight another day. A comparison between helmet mounted equipment capabilities and the mass, c of g and other implications is given.

INTRODUCTION

From the earliest moments in the history of aviation the development of the flying helmet has followed the development of aircraft's flight capabilities. Originally the helmet was needed to protect the pilot from the elements, and goggles were worn to protect the pilot against the slipstream. Later, when morse and airborne radio became available for a pilot to communicate with the ground and other aircraft in his formation, the helmet was the obvious location for the earpieces and the microphone. Noise attenuation could also be incorporated. Mission effectiveness began to rely on head mounted equipment. As aircraft operating heights increased, the use of an oxygen mask enabled pilots to operate efficiently at altitude. With the arrival of jet powered aircraft came the requirement to escape at high speed. The invention of the ejection seat placed on helmet designers the requirement to protect the pilot from airblast, canopy debris and head injuries. By this time the cloth cap weighing a few grammes had developed into the 'bone-dome' with integral visors with an overall weight of over 2 kilogrammes.

The weight of a British Mk 4B helmet, which is the current UK flying helmet, is approximately 2.25 kg including the oxygen mask. It is not uncommon for pilots to pull 7 g in combat and peak loads in excess of this figure are by no means unusual and are likely to increase as aircraft become more agile. It is recognised that any additions to the helmet which might improve his operational effectiveness may well increase pilot fatigue at best or increase the risk of more permanent damage. Other papers in the conference proceedings address in a variety of different ways, the effects the demanding environment places on the pilot's neck and how the epidemiology of neck injuries may be better studied and understood. This paper attempts to explain the operational necessity for imposing further physiological demands on the pilot and how equipment designers are trying to meet the opposing objectives of a good man-machine interface and equipment which will not lead to temporary or permanent injury to the pilot. An overview of developing technologies is given and an indication of likely future trends offered. The views expressed in this paper are those of the author and do not necessarily represent those of the RAE or MOD.

PERFORMANCE REQUIREMENTS

In this section an idealised cardinal points performance specification of requirements is suggested for a generic helmet mounted display/sighting device. Without even considering the actual function of the device, the physical characteristics place severe constraints on the designer.

Mass	Minimal (zero?)
Inertia	Minimal (zero?)
Centre of gravity	Close to (or behind) the c of g of the head
Volume	It must not impair pilot's normal head movement envelope in the cockpit
Cabling	Minimum number and bulk. Interconnection via quick release
Power	Minimal. EHT generated locally
Ejection safety	Must not increase the risk of injury to the pilot on ejection
Eye relief	Consistent with ejection, donning and doffing the helmet
Obstruction	Must not restrict the pilot's vision
Field of view	Must be adequate for the task

Visor	Compatible with the normal operation of the protective visor and sun visor
Oxygen mask	Compatible with oxygen mask
Brightne	Capable of use in all light levels
Colour	Compatible with other cockpit devices
Focus	The image must be collimated
Exit pupil	Must allow for normal helmet slippage
Robustness	Must be pilot-proof

A performance specification for a particular device would include many other elements in addition to the above factors. The net effect is a confused matrix of compromises which has resulted in the many variations in the hardware solutions from different manufacturers. None of them fully meet the above specification but all provide the pilot with something of benefit to his mission. The difficulty is deciding on the correct blend of compromises for a particular device. It is not the intention of this paper to review the different manufacturers' approaches to this difficult problem, but in the following paragraphs an overview of a variety of devices will be given in the course of explaining the operational reasons for their need.

THE NEED FOR HELMET MOUNTED DEVICES

The design of the modern flying helmet as typified by the Mk 4 helmet shown in Fig 1 has remained fundamentally unchanged for a considerable period of time. However, as aircraft performance improves and weapon and avionic systems become more capable, the pilot's workload increases and he has less time to make crucial decisions and carry out vital actions. It is essential that the pilot can operate his aircraft and its systems to the boundaries of the flight envelope with efficiency during combat. It is vital that he maximises the time spent looking outside the cockpit. These stringent and perhaps conflicting requirements are forcing designers to locate data in front of the pilot's eyes no matter where he is looking.

The requirements of helmet design inevitably lead to compromises and trade-offs. Do we design for maximum mission effectiveness at the expense of pilot safety? When does safety in terms of protection against the environment come into conflict with safety in terms of enabling him to fight better and reduce his chances of getting shot down? This paper offers no solution to that dilemma. The compromise between mission effectiveness and risk to the pilot is not easily defined.

On the training side, is it acceptable for a certain percentage of pilots to suffer temporary injury as a result of training with a certain item of equipment which gives him a significant operational advantage? It could be that by limiting the amount of g pulled when using that equipment, the risk of even temporary injury could be minimised, although at the expense of reducing operational realism and hence, in peacetime, losing training effectiveness. The more capable the aircraft and its mission avionics and the more effective weapon systems become inevitably results in military operators exploiting their newly acquired advantage. To do this effectively in wartime, peacetime training must be realistic. In training circles, the axiom 'Train like we fight and fight like we train' has become a meaningful cliché. The situation is however, far more complex and is beyond the scope of this paper.

CURRENT DEVELOPMENTS IN HELMET MOUNTED EQUIPMENT

This section examines some of the items of helmet mounted equipment which have been and are being developed. Most of the items are for use in fixed wing aircraft which is probably more relevant to the theme of the symposium. However where equipment specific to helicopter operations is discussed it will be noted as such. The list is not comprehensive but will give a good overview of likely trends as well as explaining the operational needs.

VISUAL HELMET MOUNTED DEVICES

Helmet position sensors

To make best use of helmet sights and displays, it is essential to measure the pilot's head position. This data can then be used by the aircraft avionic/weapon system in a variety of ways as will be described later. There are various devices in existence, the one which has perhaps gained widest acceptance operates on electro-magnetic principles. A radiator fitted to the aircraft structure, usually the canopy, emits phased radiations which are detected by orthogonally mounted coils in a detector unit on the helmet. This incidentally requires electrical connections as well as the sight, but only weighs a few grammes and occupies a volume of less than 5 cc. The signals generated by the coils are processed to produce an accurate measure of the helmet position, and hence the pilot's line of sight.

Helmet mounted sights

Probably the first item to be appended to the modern flying helmet was a sight. Designed to provide the pilot with an aiming mark which he overlays on the target, the device has a fairly narrow field of view.

The pilot's line of sight information may be fed to the radar or weapon system of the air craft. This technique is employed in the AH 64 Apache helicopter which has a chin mounted gun which can be trained in elevation and azimuth by the pilot pointing his head at the target.

In the case of an aircraft equipped with a missile employing infrared guidance, the missile seeker head is slaved to pilot's line of sight to enable the missile to lock onto the target. When lock-on has been achieved the missile is fired and the pilot's attention can be turned elsewhere. The operational advantage is that the pilot can effectively increase the manoeuvrability of his aircraft and thereby improve the acquisition capability of the weapon system. The important point to note is that there is no need to fly the aircraft so accurately. The operational advantages are obvious. Fig 2 illustrates the concept.

Taking the idea a stage further, it enables the pilot to illuminate the target with his radar by maintaining visual contact. The radar reflections then control a radar guided missile onto the target.

A system similar to the one described above, known as VTAS (Visual Target Acquisition System), first flew in service in F4 Phantoms in the early 70s and was used in conjunction with Sparrow and Sidewinder missiles. Fig 3 shows a helmet fitted with VTAS.

The type of symbology which will enable the tasks described above to be achieved can be quite simple and hence can be produced by a fixed reticle. Fig 4 shows a typical arrangement. The format consists of an aiming mark at the centre of a circle which defines the seeker head field of view. Markers indicating the state of the seeker, caged or uncaged, lock-on, etc, are positioned at the extreme of the sight field of view. If radar is used further markers showing the state of the radar can be included. Each symbol is illuminated as appropriate to the status of the engagement.

As far as the optical design is concerned, the light source can be filament bulbs or, more usually, an array of light emitting diodes. The image produced by the light passing through the reticle is collimated and reflected off the visor by a series of lenses and mirrors or, more usually, a prism. The design of these elements is critical to the performance of the device. Fig 5 shows a typical arrangement.

The optical design constraints can be summarised by the formula below where D_E is the diameter of the exit pupil, D_C that of the collimating lens, and d_R the distance from the lens to the eye. For a field of view 2A

$$D_C = 2d_R \tan A + D_E .$$

For a wide field of view (large A), a large exit pupil (D_E), and good eye relief (d_R), a large diameter collimating device (large D_C) is the result. However, to maintain a low weight and reduce obscuration, a small collimator is desirable. This is the basic dilemma which results in the range of products available from the manufacturers.

In order to obtain a bright image, the visor must be an efficient reflecting surface in the area where the sight image is seen. It is important that the transparency of the visor is not reduced or the pilot's view of the world becomes attenuated. To overcome this undesirable characteristic, narrow band or dichroic filter material can be used which is designed to reflect the wavelength of the sight light source and hence transmit all other wavelengths. The net result is a bright sight image and little discernible change to the scene brightness.

Once this concept had been shown to work, it was realised that by introducing new elements into the format, a helmet mounted sight could be utilised for other functions. For example, by introducing cueing marks, the pilot could be shown the direction an approaching target. A possible format for such a sight is illustrated at Fig 6. The number of different elements is now getting quite large and packaging becomes quite difficult within the constraints of a head mounted system.

One way of overcoming this is by using an array of light emitting diodes to form a matrix. Such a device is shown at Figs 7 and 8. The problem with this technology is that the number of elements in the matrix is limited and therefore the definition of the display leaves much to be desired. Vertical and horizontal lines can be shown with ease, but lines at angles exhibit a staircase effect as shown. However, an advantage of matrix displays is their ability to show alphanumerics although confusion of similar characters can occur if the number of elements in the matrix is limited, so the formats must be carefully designed. The answer is to produce arrays with smaller elements to increase the resolution of the display and much work in this area is being carried out, but so far a matrix with sufficient resolution has yet to be perfected for helmet sight application.

Helmet mounted displays

The advantage of the types of devices so far described is that they can be designed to be reasonably small and lightweight. Most of the driving electronics, power supplies etc can be mounted off the man, and the cabling requirements are not too severe. However, the trade-off is that they restrict the information available to the pilot. To provide full flexibility of display, the cathode ray tube must be utilised. This ubiquitous device has been used in many ways in aircraft, from radar screens of the 1940s to the present day 'glass cockpits' where the primary instrument panel consists of several colour CRTs. It is no accident that cockpits have developed this way. CRTs allow full flexibility of display format, giving a multifunction capability to the surface. Alphanumerics and symbols can be produced with high definition and brightness using cursive or stroke writing techniques, or if imagery is required to be displayed from an on-board sensor such as television or infrared, raster techniques can be utilised.

The early tentative steps to incorporate a CRT on a helmet were fraught with difficulties, the main being the development of a sufficiently small, rugged tube with high enough brightness and resolution. Before head mounted CRT devices are discussed an explanation of the operational need should be given.

The flexibility of format has already been mentioned. The pilot of the modern combat aircraft is normally equipped with a suite of head down or panel mounted instruments, and in the case of more modern aircraft these are CRT based. His primary instrument is the head up display, or HUD, and from this display surface he will acquire primary flight information together with secondary, or mission, data.

The display is switchable for a variety of modes depending upon the phase of the mission. Most HUDs have a general or navigation mode (as shown at Fig 9), a bombing mode, an air combat mode etc. If moding was not employed, the display would be too cluttered thus denying the pilot a clear direct view through the combiner. A typical figure for the HUD field of view is 25°. If the pilot looks outside this relatively small area he deprives himself of all flight and mission data. In air combat and air to ground operations this could severely compromise his operational capability.

The ability to keep an awareness of the situation all around the aircraft is the key to mission success and survival in a hostile environment. The phrase 'situational awareness' applies to all aspects of military operations, whether low level ground attack or air combat at higher altitudes. The continuous availability of data concerning the aircraft and its systems while the pilot scans the sky is essential to maintaining situational awareness. A helmet pointing system could also assist in the low level navigation task by defining off track way-points, or targets of opportunity, thereby reducing the pilot's cockpit workload.

It has been established for many years that to form an effective defence in Western Europe our pilots must capable of flying throughout the whole 24 hour period. The development of electro-optical sensors such as low light TV, Night Vision Goggles (NVG) and Forward Looking Infrared (FLIR) imaging systems has provided the means to accomplish this. However, if we are to operate by night as effectively as by day we must provide the pilot with the same visual information by night as is available to him by day.

The above operational requirements demand much from the helmet mounted equipment. The need for maximum flexibility of format and the ability to display sensor imagery drives us towards CRT based systems.

Table 1 gives an indication of the functions that might be required by a pilot on a helmet device and illustrates the limitations of a simple sight and a matrix display compared with cursive and raster CRT displays. It should not be implied that all the functions listed under a particular device could be shown at the same time.

Table 1

Display capabilities

Function	Fixed format sight	Matrix display	Cursive CRT	Raster CRT
LoS Wpn aiming	√	√	√	√
Seeker fov	√	poorly	√	√
Radar scan	√	poorly	√	√
Seeker lock	√	√	√	√
Radar lock	√	√	√	√
Flight data		limited	√	√
Systems data			√	√
Cueing	poorly	poorly	√	√
Reverse cueing	poorly	poorly	√	√
EO imagery				√

LoS = line of sight

The above table, which is not functionally complete for reasons of classification, clearly indicates the advantages of CRTs. Of course there are differences between raster and cursive technology apart from the latter not having an imaging capability, but this is probably beyond the scope of this paper. Having seen the theoretical advantages of a CRT based display system, some of the hardware offering those advantages will now be described.

One of the first systems to be assessed is illustrated in Fig 10. This was a monocular raster based CRT and consisted of the tube in its housing together with the optics to collimate the display which the pilot viewed on a reflector mounted in front of his eye. The device was heavy, the asymmetric c of g and the drag of the cables gave problems even in a simulator assessment. It seems likely that the device would give severe problems under g and during ejection from a fast jet. However, the device did provide experience in using helmet mounted equipment in the cockpit.

The device was part of an experimental system for helicopters developed at Farnborough known as REDOWL (Remote Eyes in the Dark Operating Without Light). In essence the pilot was equipped with a helmet mounted CRT which displayed the imagery from an EO sensor steered in the direction of his line of sight. The EO sensor was mounted under the chin of the helicopter.

It is apparent that the weight distribution is asymmetric, the volume is large, and each eye is provided with totally different information.

Trials involving a fixed wing aircraft have also taken place at Farnborough but utilising helmet mounted equipment loaned from the United States. The HIPSEOS programme (Helmet Imaging and Pointing Systems for Electro-Optical Seekers) provided a useful fixed wing experience of HMDs.The experience of operating with devices like these provided an insight to the types of problems that had to be overcome if helmet mounted display devices were to become operationally viable. The programmes not only provided technical system performance data, but introduced some of the human factors which must be considered.

The display is essentially a miniature head up display (HUD) and consists of the tube and its focussing and collimating lenses in a rigid protective tubular assembly. The image is reflected off a mirrored surface to the pilot's eye. The requirements demand a large field of view together with a large exit pupil to allow for movement of the helmet on the head. These requirements lead to the quite cumbersome designs.

Eye point of regard

Mention was made earlier of the requirement for a head position sensing system and its use by the pilot pointing his head to aim a weapon. This does lead to an unnatural mode of operating, particularly when tracking a moving target when the tendency is to follow it with a combination of head and eye movements. It is possible to measure where the eye is pointing and several devices are available although mainly for workload study applications. They operate in most instances by monitoring the corneal reflections of a spot of non-visible infrared light. At the moment the technology needs to be improved for in service applications but increased activity in this area is underway in various countries. No doubt when a mature design is available, its potential for improved aiming accuracy will be fully exploited.

Night vision goggles

Perhaps one of the earlier items to be bolted onto the flying helmet in recent times was the Night Vision Goggles (NVGs). Due to their relative cheapness and ease of installation they found immediate applications in military helicopter operations. Fig 11 illustrates a typical

installation. Energised by batteries carried on the back of the helmet (counterbalancing the goggles) the device consists of a pair of image intensifiers configured binocularly. They are fitted with an objective lens and an eye piece to focus the images individually. The important point to note is that the pilot can see an enhanced image of the visible spectrum wherever he looks. In addition he can see round the goggles to view his cockpit displays. The mounting arrangement allows for vertical, horizontal and interpupillary adjustment, and they can be parked out of sight by swivelling them on the helmet. A quick release device facilitates attachment and detachment from the helmet.

In order that the NVGs are not overloaded by the cockpit lighting, special lighting and filtering techniques must be used.

Originally used in the less demanding helicopter environment, they weighed approximately 800 gm including mounting brackets and battery pack. It was found that they could be worn for considerable time without undue discomfort, other than tired eyes perhaps. However, they soon found application to fast jet operations where a boresighted infrared image on a head up display complemented by the night vision goggles giving a wide total field of view of the enhanced visual spectrum provided a useful facility.

They obviously presented an ejection hazard and the weight and inertia was a significant addition. Automatic separation in the event of ejection is available.

Human factors

Most devices available so far have been monocular, primarily in the interests of minimising weight and complexity. There is evidence to suggest that binocular rivalry has caused problems to some pilots. Most people have a dominant eye, the tendency being for right handed to be 'right eyed'. All installations known to the author cater for the right eyed pilots.

The transferral of attention from the display to the real world and the additional task the brain has in disseminating the information provided from the different visual stimulae from each eye may well contribute to excessive workload or even disorientation.

There are several advantages of providing a binocular device. Both eyes are provided with the same stimulus which is the natural way the brain receives visual information. Stereoscopy provides the pilot with natural ranging cues, although true replication is not possible unless the aircraft is equipped with a dual camera installation. A binocular device does however incur optical design problems and doubles the weight on the head.

With a head slaved sensor it is important that the pilot is given the correct image for his direction of look, not only statically but dynamically. The turret in which the sensor is mounted must respond to rapid head movements otherwise the chances of disorienting the pilot are increased. There are more subtle affects also, relating to the dynamic characteristics of a scanned image when viewed binocularly.

If the pilot is provided with airframe referenced information, for example aircraft attitude or heading, on his helmet mounted display, great care must be taken in the way the information is presented otherwise it could lead to disorientation when he moves his head off boresight.

These problems indicate that there are many human factors considerations which need to be resolved.

NON-VISUAL HELMET MOUNTED DEVICES

The paper has so far concentrated on helmet mounted devices of a visual nature. This section deals with non-visual devices which are located on the helmet.

Active Noise Reduction (ANR)

The level of noise in modern military cockpits is in general increasing. A method of improving the situation is by measuring the noise in the earpiece of the helmet and reintroducing an inverted signal to achieve cancellation. As far as the helmet is concerned the only additional component is a miniature microphone mounted within the earpiece.

Adaptive Noise Cancellation ([illegible])

This is aimed at reducing noise on the pilot's telephone line using digital filtering and correlation techniques. As with ANR, additional weigh should be minimal but additional leads to the helmet would be needed.

Directional audio cueing

There is a possibility that by using stereophonic earpieces in the helmet, a pilot could be given an audio warning of the direction of a threat. Considerable work on the human factors aspects is needed before this technology matures.

Nuclear, biological and chemical warface

Hoods designed to protect the pilot against NBC threats are invariably uncomfortable and restrict the pilot's head mobility and his visual field. The operational implications of wearing them are not trivial. Head mounted equipment should therefore be designed with NBC protection in mind. With an enclosed head assembly, two factors grow in significance. The pilot must be provided with an automatic visor demisting system and it seems likely that some form of head cooling arrangement should be incorporated.

TRENDS IN HELMET MOUNTED EQUIPMENT

The major problems associated with these devices, especially those offering visual information, are the size, weight and the fact that they are mounted in front of and in close proximity to the pilot's face. Their location can not be changed until such times as a means of directly inputting data to a pilot's brain can be developed. Faced with the reality of where the devices must be placed, the component industry must be encouraged to develop smaller, more effective building blocks.

Industry is improving the efficiency of LEDs to provide higher brightness with less heat dissipation. Improved resolution could bring the matrix display to the level necessary for portraying EO imagery. Integrated logic/drivers capable of handling the peak powers involved will enable the display and electronics to be more reliable and reduce the weight on the helmet.

The need for a more robust, higher brightness, better resolution, smaller and lighter CRT remains. Industry is actively pursuing miniaturisation of the devices. Electronics can be packaged smaller and the option of mounting the drive electronics and EHT units within the helmet shell is now a possibility with obvious advantages for the system integrity. Fig 12 gives an example of what can be achieved.

However, all devices to date require optical components to collimate and combine the symbology or imagery with the direct view of the outside world. This can take the form of a glass combiner in front of the pilot's eyes, which is undesirable in the event of a crash or ejection, and imposes some occultation of the pilot's view. An alternative is a reflective patch on the visor. This is a better device from the flight safety point of view but places restraints on the optical design of the system. One of the features required of a helmet mounted display is a large exit pupil or port-hole through which the pilot can view the display.

The most promising technology which could provide the solution to both these problems is holography. A holographic visor could provide a very efficient method of combining the display with the outside world view. In essence a hologram may be regarded as having optical power at a specific spectral frequency but zero power at all other frequencies. A holographic visor therefore reduces the optical elements on the helmet, significantly reducing the weight, could provide a very wide field of view, and a large exit pupil. However, at the moment the problems of developing a hologram compatible with a curved polycarbonate visor still need to be solved.

To capitalise on the above developments it is important that the pilot can aim the device with accuracy. Apart from the pilot's difficult tracking task, the weapon system must be given accurate information of where the pilot is looking. An eye point of regard device would assist greatly in both of these functions.

NVG developments have tended to be directed towards improving the image intensification, autogain control, and filtering techniques to improve the perceived image. Recently, more attention is being paid towards reducing the mass and also the moment of inertia by minimising the weight of the objective lens and integrating lithium batteries within the goggles.

The trend until today has been to design a helmet which meets the basic protection, life support, and communication requirements, and add on the new devices in a seemingly *ad hoc* manner. The complex interactive nature of the devices we are dealing with does not allow us to pursue this approach any further. Specifications for the weapons and avionic systems of aircraft are being written in a 'top down' manner. Technology has now placed the helmet firmly in the 'systems' area and future designs should reflect this.

The trend for modern helmets and helmet mounted equipment must be to design in an integrated manner. The shape of the shell must not only be designed with protection, life support and communication in mind, but also with the mission equipment high on the list. Work is underway

to develop a helmet designed from the outset with integral NVGs, HMD and HMS in addition to standard protection and life support capabilities.

Integration must be functional as well as physical. If the functions of two or more devices can be integrated into a single new device then it makes sense from all aspects to develop the new device. Most of the man-mounted equipments described in this paper require electrical connections which must be routed via the seat to the aircraft systems. It is clear that the number of leads must be kept to a minimum. A method of achieving this is by designing equipment to operate on standardised power supply rails. There seems to be scope for a form of digital data bus to the helmet for such things as a sight and a head/eye pointing system.

CONCLUSIONS

Referring back to the list of performance requirements, the reader will see the physical aspects to be considered in designing an HMD. The trend must be towards reducing the size and mass of the devices, whilst at the same time improving their optical characteristics. Higher brightness, better resolution and contrast CRTs contained in smaller volumes is the goal to be achieved.

To overcome the problems of operating a helmet mounted display in a high g environment without compromising a successful ejection, designers are beginning to produce hardware capable of meeting many of the requirements. Recent developments have shown that it is no longer viable from a design or operating point of view to consider the helmet and helmet mounted devices as separate entities. Work is underway to produce an 'integrated helmet' which is designed from the outset to provide protection, communication, sensor imagery and flight/systems data to the pilot. As well as a better helmet system design, the likely outcome will be a lighter helmet assembly. Specifications are being written which call for an inclusive helmet mass of the order of 1.5 kg. It will be interesting to see the outcome.

The time has come when experts from heretofore separate disciplines must combine their talents. At the concept stage of a man mounted device or an aircraft project the optical designer, the weapons systems engineer, the mission avionics system designer, the communications specialist, the human factors engineer, the aircrew equipment assembly designer and the aeromedical specialist must all collaborate and combine their talents if the final outcome is to be a success.

Fig 1 Mk 4 flying helmet

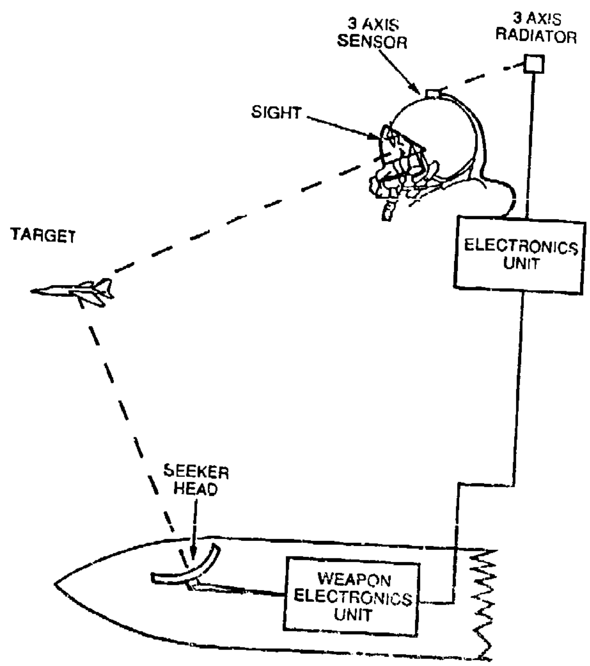

Fig 2 Off-boresight weapon aiming

Fig 3 Visual target acquisition system (VTAS)

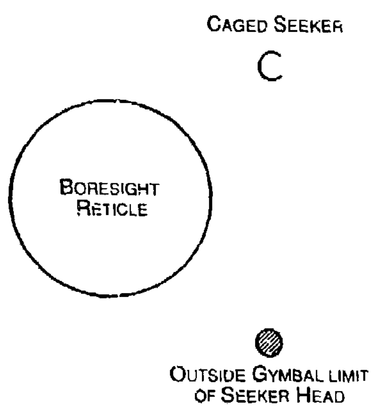

Fig 4 Basic helmet mounted sight symbology

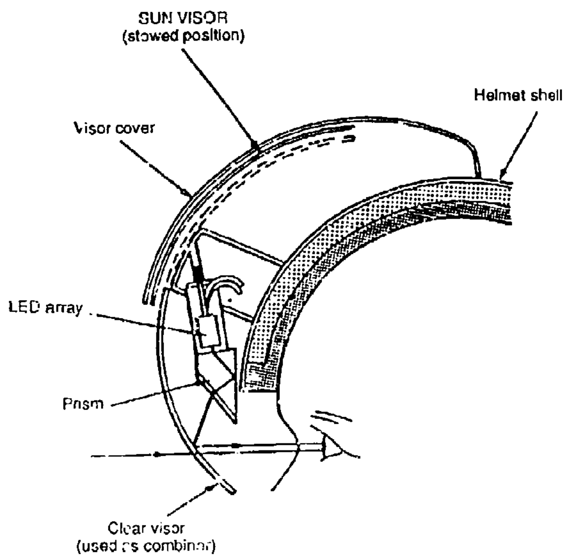

Fig 5 Helmet mounted sight

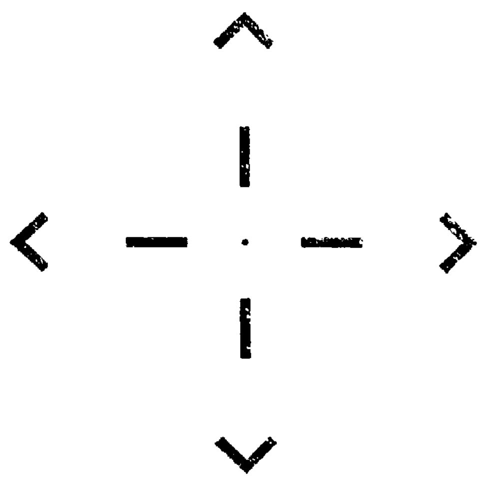

Fig 6 Helmet mounted sight cueing symbology

Fig 7 Helmet mounted matrix display

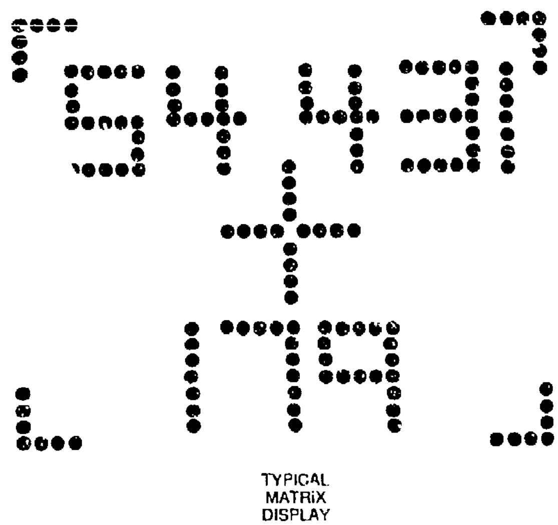

Fig 8 Helmet mounted matrix display symbology

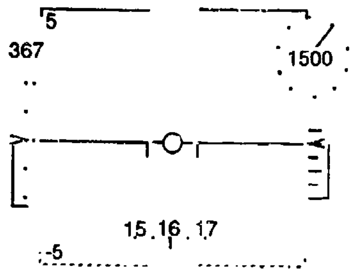

Fig 9 Typical head up display format

Fig 10 Helmet mounted CRT display

Fig 11 Night vision goggles

Fig 12 A modern helmet mounted CRT display

PREVALENCE OF G-INDUCED CERVICAL INJURY IN U.S. AIR FORCE PILOTS

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SUMMARY

Pilots of high performance aircraft are frequently exposed to significant accelerative forces; the effect of this exposure on the cervical vertebral column is an unquantified clinical and epidemiological problem. This report presents the findings from a period prevalence study of acute neck injury secondary to high G forces in U.S. Air Force pilots. A sample of 437 pilots of three different fighter aircraft was surveyed, and the data is stratified and analyzed to test the strength of association of injury prevalence with pilot age, type of aircraft, and type of flying environment. Results indicate that minor injury is common in these pilots, and that higher aircraft performance is associated with increased injury prevalence. Increased age is associated with increased prevalence of major cervical injuries. Preventive strategies may be helpful in reducing injury frequency and in avoiding serious injuries.

INTRODUCTION

The performance capabilities of current high-performance fighter aircraft have greatly expanded since the introduction of the F-15, F-16 and F-16. One of the most significant advances has been in the capability to sustain higher positive gravitational forces for longer periods of time and more frequently in all flight environments. This capability exposes the pilot to a significantly more severe occupational stressor than that of previous fighter-type aircraft. Several studies have assessed potential adverse health effects from this high "G" environment on various organ systems (1,2,3,4,5,6,7,8). Most of these studies have focused on the neurologic, cardiovascular and respiratory systems, with particular emphasis on fatigue, performance deterioration, vision effects, loss of consciousness, coronary blood flow, and cardiac arrhythmias. However, few studies have addressed the short or long term effects of brief or sustained high G forces on the musculoskeletal system (9,10,11). There are frequent pilot reports of neck strain yet little data exist on the possible pathologic effects to the skeletal system of humans from such exposures.

Several subclinical musculoskeletal conditions (Appendix 1), predominantly spinal conditions, were identified in a U.S. Air Force working group in 1979 (12); it was postulated that these conditions would be aggravated by high sustained +G. Various recommendations were made by this workshop for added screening exams to detect these conditions, and to revise the physical standards for entry into pilot training for applicants with these musculoskeletal conditions. However, there is insufficient data to demonstrate that these preexisting musculoskeletal conditions have resulted in increased injury prevalence in pilots of advanced aircraft, or that those without such conditions are less likely to develop spinal problems.

The gravitational forces exerted on the pilot's body may be applied in any of three axes: 1) the "fore-aft" or x-axis, denoted Gx; 2) the "left-right" or y-axis, denoted Gy; and 3) the "head-foot" or z-axis, denoted Gz. In the high G environment, the x- and y-axes are not significantly stressed; the Gz axis however experiences high forces in the positive direction (+Gz, i.e. from head to foot), up to +9G's or greater, with much smaller forces in the -Gz direction (rarely exceeding -2G's). High sustained G is defined as greater than 6 G's for 15 or more seconds. Advanced fighter aircraft are now able to achieve and sustain high G maneuvers for lengthy periods under certain conditions, and structural integrity may not be a consideration until levels of +9G's or greater are achieved. This high G environment is experienced in actual combat, in frequent training missions for simulated aerial combat, and during advanced aircraft handling maneuvers.

The portion of the musculoskeletal system subjected to the most severe stress is the vertebral column. Column strength is reduced with flexion or torsion movements (9); thus the neck is particularly susceptible to injury as these cervical vertebrae can achieve much greater departures from vertical alignment than can lower segments. High G forces combined with frequent turning and tilting of the head and neck increase the potential stresses to the cervical vertebrae. The weight of protective headgear and oxygen equipment add to this stress. Seat back angle has an effect in the dispersal of +G forces into the Gx axis as seat back incline is increased. However, many pilots report that they lean forward, particularly in flexion of the cervical spine, during high G maneuvering to enhance their visual search or to maintain visual sight of an attacker, thus negating the potential benefit of the increased seat angle to the cervical vertebrae. New life support systems are reported to reduce the required straining effort by 50% to maintain consciousness at high +G; with further advances in the support of the pilot's cardiovascular, pulmonary, and neurologic systems, the cervical vertebrae may become the high G "weak link" in the human system.

There are frequent anecdotal reports of acute neck injury in the fighter pilot community. Some pilots report a higher frequency of neck strain in advanced fighter aircraft when compared with older aircraft. However, there are no published reports of the actual prevalence of "everyday" neck injuries, nor of their character, quality, duration or sequelae. Neck injury may range from a mild dull ache to pain and spasm, or may present as a debilitating injury with sensory and/or motor deficits. Muscle strain may present as localized neck tenderness, or may radiate from the occiput to the shoulders or the area between the scapulae. Spasm may or may not be present. Cervical nerve root injury may present as a dull ache or pain in the neck, shoulder or arm, or may be described as a numbness or tingling in the

distribution of the affected nerve, extending distally to the hands and fingers. In severe cases reflexes may be diminished and there may be impairment of motor function of the affected upper extremity.

Certain serious injuries can be documented by various radiological methods. These lesions include fractures of the vertebral body or arch and herniated intervertebral disks. Milder injuries cannot always be demonstrated by diagnostic methods and are referred to as acute cervical syndromes. These are defined as any of several entities caused by irritation or compression of the cervical nerve roots. Several types of such syndromes are recognized: 1) muscle pain or tenderness with or without radiation into the back or shoulders; 2) muscle spasm; 3) torticollis--a more severe strain with contraction of the cervical muscles producing a twisting of the neck and a resultant unnatural head position; 4) sensory deficits (paresthesias or dysesthesias) in the distribution of the affected nerve root; and 5) motor deficits, with decreased deep tendon reflexes or frank impairment of coordination, dexterity or movement (13,14,15). Paresthesias refer to the symptoms of burning, numbness, tingling or prickling. Dysesthesia refers to any sensory impairment, but especially touch.

Many pilots fail to seek medical attention for such injuries unless there is an impairment of flying abilities. Thus, many cases are not reported to the medical community. There are only a few documented cases of cervical spine injury with either vertebral fractures, herniated nucleus pulposus, or ligamentous tear secondary to high G exposure (16). It is probably still too early to ascertain whether or not chronic G exposure leads to any long-term disability.

Thus, neck injury and its sequelae remain an unquantified clinical and epidemiological problem in pilots exposed to high G forces. The purpose of this study is to describe the period prevalence of this occupational injury. Types of injuries are described and classified. Comparisons are made between the prevalence of injury by type aircraft, pilot age, and type of flying. The study may facilitate future assessment of occupational risk, and may direct research towards improved G protection for the musculoskeletal system of exposed pilots. This data may have implications for human factors design of even more advanced future combat fighter aircraft. Can man tolerate stresses in the Gz axis of +10g or more? What is the operational tolerance of the human cervical spine? These questions need to be addressed; this study may be a useful step towards finding the answers.

METHODS

An anonymous survey questionnaire was utilized to collect the data for this descriptive period prevalence study. First the survey form was described and explained uniformly to all participating pilots, then the questionnaires were distributed and collected after ample time was afforded to complete the form. Anonymity and confidentiality were highly stressed. Specific data collected included the number and type of neck injury by time period, pilot age and flying experience by type aircraft, total flying hours, crew position, and type of flying currently performed (operational or training environment). History of any prior traumatic vertebral injury was also identified.

Presence or absence of neck injury is the dependent variable, and is quantified as the numerator by the number of pilots with at least one acute cervical injury in the time period specified. The denominator is the population at risk, i.e. all G exposed pilots for each of the sample strata. Injury may be further subdivided into "major" and "minor", with the more severe injuries presenting with distal upper extremity symptoms. In addition, "major" injuries could be expected to potentially compromise flying safety by adversely affecting aircraft control.

The data do not include individuals with a prior history of vertebral fractures or herniated disks. Those with chronic cervical syndromes were also excluded from the analysis. The data allows for quantification of the number of cases in each period, as some pilots have more than one injury per period; this may warrant further study into types of pilots who seem more susceptible to recurrent injury than others, however it is not addressed in this study.

The potential for survey bias exists, and several factors were considered (17). Anonymity was very highly stressed, and the questionnaire was kept simple, straightforward, and brief. The selection of air bases for the survey is considered unbiased, as the type of flying at each base with similar aircraft is considered to be essentially identical. Sex is not a consideration since all pilots surveyed were male. Headgear is not a factor, as all U.S. Air Force pilots wear similar helmets and oxygen equipment of very similar weights. All pilots entering into Air Force pilot training meet the same physical examination and screening standards, and selection for type of aircraft is not dependent upon physical criteria. Selection is based on academic and flying skills performance in pilot training, and these are not considered to be biases that might affect future potential for acute cervical injuries. Recall bias was minimized due to the type of data requested on the questionnaire, and due to the short recall intervals addressed. A pilot may have difficulty recalling the exact number of acute injuries over a specified time period, but the relevant finding was <u>any</u> neck injury in the period studied.

The period prevalence ratio is determined by dividing the number of current cases in the period by the number at risk for each sample (18). A current case is defined as a pilot who had at least one neck injury secondary to high G force exposure in the time period. Populations sampled and analyzed included pilots from both an operational and a training wing for each of three aircraft (F-5, F-15, and F-16). Expected number of pilots available for survey at each wing ranged from 50-80 for each type aircraft; actual numbers sampled exceeded this prediction, ranging from 58 to 121 pilots at the various wings.

Data for the 3-month periods are first analyzed using the CATMOD Procedure of the Statistical Analysis System (SAS), with Type of Aircraft, Age Group, and Type of Flying as the independent variables, and Severity of Injury as the dependent variable (19). These data are then arranged into RXC tables for the various sample strata, then testing for significance with Chi square tests with (R-1)(C-1) degrees of freedom (20). Significance level is set at $p < 0.05$. Data for a 3-month period was selected for analysis

for two reasons: 1) recall of injury experience over this time period should be reasonably accurate; and 2) three months of flying is likely to assure frequent exposure to high G missions.

G exposure is a quantity that must be considered in assessing occupational risk. Duration and frequency of exposure to the various G levels differs among the aircraft and with the type of flying encountered. For example, the operational flying environment may involve more high G missions than does the training environment. Experienced pilots may use less G in their air combat maneuvers than pilots with fewer hours of training. Gillingham et al (21) provided one analysis comparing the G-environments of four fighter aircraft. His study demonstrated that the F-16 tends to expose the pilot to G forces more frequently and to a higher degree than the F-15, and the F-15 moreso than the F-5.

RESULTS AND DISCUSSION

According to data from the Air Force Military Personnel Center, at the time of this survey there were 3125 active pilots in the F-5, F-15, and F-16. The survey sample size of 437 pilots thus comprises approximately 14% of the population. At the three bases visited, the total number of assigned pilots in the three aircraft was 590; thus the 437 in the sample constituted 74% of all pilots assigned, and an even higher percentage of those available, as some pilots were on leave or performing other required duties.

All pilots sampled in the survey were male The mean age for pilots surveyed was 31.5 years (range 23-47), with mean total flying hours of 1,69. range 180-4,700). In the aircraft currently flown, the mean time in the aircraft was 2.7 years and 515 flying hours (range 0.1-12 years, 10-1,850 flying hours). Pilots at training bases were older and had more total flying hours than those at operational flying bases. Period prevalence ratios are shown in Table I. The ratio shows the percentage of all pilots who experienced at least one acute neck injury secondary to high G forces in the period listed. Also shown are the prevalence ratios for major neck injuries in the same periods.

TABLE I. PERIOD PREVALENCE RATIOS (ALL PILOTS, ALL AIRCRAFT; n = 437)

Time Period	Past Month	Past 3 Months	Past Year
Any Neck Injury	30.6%	50.6%	63.6%
Major Neck Injuries	4.3%	8.7%	11.2%

The complete 3 month injury data are shown in Table II for reference. Pilots with major injuries in the period may have had minor injury as well, but such individuals are listed only once in the column for major injury. Those listed as having minor injury had <u>only</u> minor injuries in the period.

TABLE II. NECK INJURY PAST 3 MONTHS, BY TYPE AIRCRAFT, AGE GROUP, TYPE OF FLYING, AND INJURY SEVERITY.

Type Aircraft	Age Group	Type of Flying	Severity of Injury: Major	Minor	None	Total
F-5	20-29	Operational	0 (0%)	1 (7%)	14 (93%)	15
		Training	0 (0)	2 (29)	5 (71)	7
	30-34	Operational	0 (0)	7 (32)	15 (68)	22
		Training	1 (8)	7 (54)	5 (38)	13
	35+	Operational	1 (11)	1 (11)	7 (78)	9
		Training	1 (11)	2 (22)	6 (67)	9
F-15	20-29	Operational	1 (2)	15 (36)	26 (62)	42
		Training	2 (5)	17 (44)	20 (51)	39
	30-34	Operational	0 (0)	9 (56)	7 (44)	16
		Training	7 (7)	16 (55)	11 (38)	29
	35+	Operational	0 (0)	8 (62)	5 (38)	13
		Training	4 (17)	9 (39)	10 (43)	23
F-16	20-29	Operational	1 (3)	22 (61)	13 (36)	36
		Training	7 (14)	23 (46)	20 (40)	50
	30-34	Operational	1 (5)	10 (45)	11 (50)	22
		Training	6 (17)	12 (34)	17 (49)	35
	35+	Operational	3 (14)	10 (48)	8 (38)	21
		Training	8 (22)	12 (33)	16 (44)	36

The categorial analysis (CATMOD) of these data gave no indication of significant 1st or 2nd order interactions among the three factors (Type of Aircraft, Age Group, Type of Flying). However, all three main effects were statistically significant. Consequently, each factor is analyzed in more detail below. In each case, analysis is viewed from two perspectives: any injury vs. no injury; major injury only vs. all others. Chi square testing is shown below tables III thru VI. In Table III, severity of injury within the past 3 months is stratified by type of aircraft. This table demonstrates a statistically significant trend in frequency (F-15 and F-16 >> F-5) and in severity (F-16 > F-15 or F-5). The data support a hypothesis that neck injury is more prevalent in the F-16 and F-15 than in the less G-capable F-5, and that major injury is more prevalent in the F-16 than in the F-15 or F-5. This is consistent with the hypothesis that as aircraft performance capability increases, so does the potential G exposure and consequently potential G-induced injury.

TABLE III. NECK INJURY, PAST 3 MONTHS, BY TYPE AIRCRAFT AND SEVERITY

	F-5	F-15	F-16	Total
Major Injury	3 (4.0%)	9 (5.6%)	26 (13.0%)	38 (8.7%)
Minor Injury	20 (26.7%)	74 (45.7%)	89 (44.5%)	183 (41.9%)
No Injury	52 (69.3%)	79 (48.7%)	85 (42.5%)	216 (49.4%)
	n1 = 75	n2 = 162	n3 = 200	n = 437

2 x 2 Chi square tests:
Major vs. Minor + No Injury: F-15/F-16, $p < 0.025$
: F-5/F-16, $p < 0.05$
: F-5/F-15, N.S.

Any Injury (Major + Minor) vs. No Injury: F-5/F-15, $p < 0.005$
: F-5/F-16, $p < 0.0005$
: F-15/F-16, N.S.

Tables IV and V show data for neck injury in the past 3 months tabulated according to injury severity; Table IV stratifies the pilots into age groups, while Table V stratifies by type of flying environment (operational or training).

TABLE IV. NECK INJURY PAST 3 MONTHS, BY SEVERITY AND AGE GROUP (N=437)

	Age: 20-29	30-34	35+
Major Injury	11 (5.8%)	10 (7.3%)	17 (15.3%)
Minor Injury	80 (42.3%)	61 (44.5%)	42 (37.8%)
No Injury	98 (51.9%)	66 (48.2%)	52 (46.8%)
	n1 = 189	n2 = 137	n3 = 111

2 x 3 Chi square tests:
Major vs. Minor + No Injury, $p < 0.025$
Any Injury (Major + Minor) vs. No Injury, N.S.

TABLE V. NECK INJURY PAST 3 MONTHS, BY SEVERITY AND TYPE OF FLYING

	Operational	Training	
Major Injury	7 (3.6%)	31 (12.9%)	
Minor Injury	83 (42.3%)	100 (41.5%)	
No Injury	106 (54.1%)	110 (45.6%)	
	n1 = 196	n2 = 241	n = 437

2 x 2 Chi square tests:
Major vs. Minor + No Injury, $p < 0.005$
Any Injury (Major + Minor) vs. No Injury, N.S. ($p < 0.10$)

The data in Tables IV and V reveal the following: 1) major neck injury appears to be more prevalent as age increases; and 2) major injury appears to be more prevalent at training (RTU) fighter bases while minor injuries are evenly distributed.

TABLE VI. "IF YOU FLEW PREVIOUS FIGHTER AIRCRAFT, HOW DOES THE CURRENT AIRCRAFT COMPARE WITH RESPECT TO G-INDUCED NECK INJURIES?"

	F-5	F-15	F-16	Total
More frequent and/or severe	9 (30%)	19 (37.3%)	69 (71.1%)	97 (54.5%)
Same frequency or severity	21 (70%)	29 (56.8%)	26 (26.8%)	76 (42.7%)
Less frequent and/or severe	0 (0%)	3 (5.9%)	2 (2.1%)	5 (2.8%)
	n1 = 30	n2 = 51	n3 = 97	n = 178

3 x 3 Chi square test for independence, $p < 0.0005$
3 x 2 testing: F-15/F-16, $p < 0.0005$
: F-5 /F-16, $p < 0.0005$
: F-5 /F-15, N.S.

A final comparison is shown in Table VI. Of the 437 pilots surveyed, 178 had previous flying experience in other fighter aircraft (F-5, F-4, A-10). The table displays the findings by type aircraft and by whether the current aircraft gave the pilot more, the same, or lesser neck injury (either by frequency, severity, or both) than the previously flown aircraft. The findings in Table VI support a hypothesis that neck injury subjectively is more frequent or more severe in the F-16 than in other fighter aircraft; very few pilots reported that current injuries are less frequent or less severe when compared with their previous aircraft.

Space for additional comments was provided on the reverse of the survey form. Several commented on the importance of frequent flying, and that a long layoff from high G exposure seemed to lead to an increase in injury susceptibility. F-16 pilots tended to add more notes than other pilots; the following are five such comments:

"I freqeuntly have sore neck when flying BFM, usually from looking back; I haven't had problems as severe since I learned not to move my head while pulling over 6-7 G's."

"I think a good warm up before flying, and not moving head above 6 G's are a key."

"I hate to fly similar F-16 vs. F-16 because of the physical demands. I worry about neck injury on all air to air sorties."

"Exercise & training, diet, rest seem most important factors in reduce/preventing injuries..."

"In this jet, you need to loosen up neck/back prior to flying high G sorties. If I do this religiously, problems are reduced greatly."

Only 81 pilots completed the space on the survey for "most likely head position at the time of injury". Of these, about half said they were moving their head under G loads, and half said they were looking back over either shoulder "checking six".

CONCLUSIONS

The main objective of this study was to determine the prevalence of acute cervical injury in pilots exposed to high G forces. Injury prevalence was expected to be higher in the advanced fighters; this hypothesis is supported by the data. The high percentages may surprise some readers, but did not surprise the author who had been at an F-15 and F-16 training base for one year as a flight surgeon. Anecdotal information suggested a common occurence of neck injury; the survey data support this impression, with nearly half of all pilots reporting some degree of neck injury in the previous 3-month period and nearly 9% reporting significant injury in this same period.

There are other significant findings in this study. The F-16 appears to induce more frequent and more severe injury than do the other aircraft; increasing age appears to place one at higher risk of major injury; and major injuries, particularly in the F-16, are more prevalent at training bases. These findings could be explained by several theories. There may in fact be a higher G-exposure in the F-16; there may be an increased susceptibility to major injury with increasing age; and an older pilot population at the training bases might explain their higher prevalence of major injury. Another possible explanation for the higher prevalence at training bases is the mix of an older more susceptible instructor population combined with a younger less experienced student population that has not yet learned how to avoid injury in the cockpit.

Long term cohort studies are needed to assess any chronic ill effects from repeated exposures. The Belgian and Dutch cervical spine X-ray screening programs and follow-up for F-16 pilots may provide some information in this regard (22,23). Could there be a degenerative effect (such as cervical arthritis) on the cervical spine from repeated G exposures, even in the absence of fracture or disk herniation? Are repeated exposures cumulative? Are we at the edge of human tolerance with present G exposures, or is there more "G-room" in which to expand without risk of long term injury? Will enhanced G-tolerance technology for maintaining consciousness (positive pressure breathing, tactical life support systems) expose the cervical spine as the weak link in the human system?

Until further research is conducted, pilots need to be made aware of the potential for injury and of the importance of preventive measures. These measures might include: 1) a modest neck exercise program; 2) neck stretching or "G warmup" in the cockpit prior to a high G mission; 3) gradual return to high G missions after a layoff; 4) minimize movement of the neck under high G loading; 5) maintain good nutrition and fly well rested, and 6) maintain good general physical condition. Design research needs to be continued to assess any preventive effect from increasing seat back angle, with Gz forces directed more into the pilot's Gx axis. Finally, cervical support systems may be essential if advanced tactical fighter aircraft are designed to be capable of sustained performance levels of +10 Gz or greater.

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Appendix 1

Musculoskeletal Conditions Thought to be Aggravated by High Sustained G

1. Cervical Degenerative Joint Disease
2. Lumbosacral Degenerative Joint Disease
3. Spondylitis
4. Spondylolysis
5. Spondylolisthesis
6. Scheuermann's Disease (Kyphosis)
7. Prominent Lordosis or Kyphosis
8. Klippel-Feil Anomaly (Congenital Short Neck)
9. Sprengel's Anomaly (Congenital High Scapula)
10. Ankylosis
11. Schmorl's Nodes
12. Hypertrophic Transverse Process L-5 Articulating with the Ileum
13. Hemivertebra
14. Spina Bifida
15. Spinal Canal Stenosis
16. L-5 Sacralization
17. Lumbarization of First Sacral Vertebrae
18. Radiological Evidence of Basal Impression
19. Cervical Ribs
20. Scoliosis
21. Intraspongy Nuclear Herniation
22. Significant Comparison or Height Loss of any Vertebral Body

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"NON-EJECTION NECK INJURIES IN HIGH PERFORMANCE AIRCRAFT"

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ABSTRACT

The potential for significant neck injuries exists in today's high performance fighter aircraft. The G-loads required to produce injury need not be excessive, nor is experience level necessarily protective. Eight cervical spine injury cases, due to or aggravated by $+G_z$ in F-15 and F-16 aircrew members are reviewed. These include two compression fractures (C_5/C_7), three left HNP's (C_{5-6}/C_{6-7}), one fracture of the spinous process (C_7), one interspinous ligament tear (C_{6-7}), and one myofascial syndrome (C_6). Mechanisms of injury and evaluation are discussed. Exercise conditioning may play an important role in prevention and protection. The role of screening x-rays and improving equipment remain as areas where further work needs to be done.

The introduction of the F-15 Eagle to the USAF Tactical Air Command in 1974, and the F-16 Fighting Falcon 5 years later, heralded a new generation of high performance fighter aircreaft with significantly enhanced performance capability. Compared to previous generation fighter aircraft such as the F-4 Phantom II, the F-16 has twice the turn rate and uses half the turn radius. This enhanced ability to produce abrupt onset high-G loads, as well as to sustain them at higher levels than previously experienced in older aircraft, presents an additional stress to the cervical spine. If the head weighs 3.5-5kg (1) with 1.8-2.2 kg of headgear added, static load equivalents of 48-65 kg are generated at $+9\ G_z$. The cervical spine is capable of sustaining axial loads of up to 91 kg without difficulty, as evidenced by native Africans who carry loads of produce on their heads on a daily basis (2).

Fighter aircraft operate in a dynamic environment that often requires a nearly constant vigil of all sectors surrounding the aircraft. Abrupt G loading in a defensive or offensive maneuver frequently applies a significant load to the cervical spine, in an other than axial direction. This, in turn, can cause loss of head control or failure of a musculoskeletal component of the cervical spine. It has been noted that flexion and extension injuries are produced at approximately 50% of the loads which cause axial compression failure (3).

The problem of neck injuries sustained within the cockpit environment is not a new one. Phillips in 1959 (4) described a student pilot who suffered an acute flexion injury to the neck during a $+9\ G_z$ emergency pullout in an AD-4. The student recovered and landed the aircraft, but several hours later presented with ataxia. The etiology was never clearly established but was thought to be a cerebellar contusion. Neck injuries have also been described from contact with the canopy in flight, resulting in fractures of multiple cervical vertebrae and incapacitation of the crewmember (5). Recently, a compression fracture of C_6 was described in a flight surgeon from the Royal Norwegian Air Force in an F-16B during an abrupt, sustained, high-G defensive maneuver (6).

CASE REPORTS

The following cases present a spectrum of non-ejection, non-impact cervical spine injuries due to $+G_z$ loading in high performance aircraft pilots. All had a negative history for ejection or prior neck injury unless stated otherwise. Most had participated in a variety of collegiate sports from football to pole-vaulting.

Case 1. Compression Fracture of C_7

Patient A was a 27 year old white male F-15 student pilot, 180 cm tall, weighing 68 kg, with a thin build. He previously had been a T-37 instructor pilot and accumulated 1,100 hours (h) total flying time. Equipment: Helmet and mask used was a HGU-26P/MBU-5P (1.9kg) fitted with Velcro pads only (not custom fit).

Mission: Neutral basic fighter maneuver (BFM) proficiency ride.

Discussion: The student was flying the 3rd engagement of a head-on neutral BFM attack on a hazy afternoon and was having difficulty with target acquisition. Visual identification of the opponent occurred late, just prior to the merge. A diving 6-G attack was initiated. The aircraft quickly oversped in the vertical, with the student rapidly approaching the floor of their altitude block. The instructor, in the other aircraft, commanded the student to perform a tighter pullout, which he initiated. However, at 9 G the student was unable to maintain his head in an upright position and it abruptly flexed to his chest where it remained through the completion of the pullout (Fig. 1). The student experienced a sharp pain in the posterior base of the neck, but was able to recover the aircraft without further difficulty. The mission was terminated and the return to base (RTB) was uneventful. Neck soreness persisted post-flight, which the student treated with an external analgesic rubbing compound. He noted no neurologic symptoms.

The following day the student flew four engagements and repeatedly experienced sharp neck pain above 4 G. He saw the flight surgeon (FS) after the mission, who diagnosed a cervical muscle strain and prescribed analgesics and muscle relaxants. The medication caused nausea and vomiting and the patient returned the following day. Osteopathic manipulation was then performed and the medication was changed. Feeling better, he was returned to flying status (RTFS) 3 days later. The next day he flew a mission to 5 G which caused significant pain and muscle spasm. C-spine x-rays showed a 22% compression fracture of C_7 (Fig. 2). A bone scan performed 7 days after the initial injury showed uptake at C_7 (Fig. 3). The patient was placed in a Philadelphia collar for 6 weeks, followed by a soft collar for 3 months. The patient no longer desired to fly high performance aircraft, and was RTFS with an Air Force Category IIB waiver (tanker, transport, bomber aircraft—no ejection seat). He has had no further problems.

Case 2. Mild Compression Fracture of C_5

Patient B was a 29 year old white male F-15 instructor pilot (IP)/Fighter Weapons School (FWS) graduate, 168 cm tall, 77 kg, with a muscular build. He had 1,150 hours total flying experience which, except for undergraduate pilot training (UPT), was all in the F-15.

Equipment: Helmet and mask used was a HGU-48P/MBU-12P (custom fit).

Mission: Defensive BFM.

Discussion: The incident occurred during the first 1 vs. 1 engagement, with the mishap IP in the lead F-15. The student was positioned in a low 5:30 o'clock position relative to the IP's aircraft (the nose of the aircraft is 12 o'clock, the tail 6 o'clock). The instructor began a 6.5 G descending left break turn away from the student, moving the student's aircraft to a high 7 o'clock position on the mishap pilot's canopy. The instructor moved his head during this turn from a right downward gaze to a left upward gaze. During this transition, at the midpoint of rotation, his neck became "locked." The instructor, not about to be defeated by a student, immediately forced his neck through this resistance, experiencing some crepitus and pain in the posterior neck. No neurologic symptoms were noted. The mishap IP flew two more engagements and repeatedly experienced pain on G loading. C-spine x-rays showed a 10% compression fracture of C_5 (Fig. 4). A bone scan showed uptake at C_5 (Fig. 5). The patient was treated with a Philadelphia collar for 8 weeks, a soft collar for 4 weeks, followed by physical therapy and neck conditioning exercises. He was RTFS 3.5 months later. He has had no further problems, and is still flying the F-15.

Case 3. C_{6-7} Interspinous Ligament Injury

Patient C was a 24 year old white male F-15 student pilot, 188 cm tall, 86 kg, with a lean build. He had a total of 300 hours flying time.

Equipment: Helmet and mask used was a HGU-26P/MBU-5P (custom fit).

Mission: Defensive BFM.

Discussion: The student was flying the fifth engagement of a defensive BFM mission. He suffered a neck strain while checking his 5 o'clock position over his right shoulder at a G load of approximately 4.5-5.5 G. He experienced mild soreness after the flight, but did not seek medical attention. He played volleyball that evening, and, while performing a right-handed overhand spike, experienced an exacerbation of his neck pain during the follow-through motion. The following morning he had further pain and spasm, with slight paresthesia in the right arm. This resolved in 48 hours. Cervical spine x-rays showed a widened C_{6-7} interspinous ligament (Fig. 6). He was treated with a Philadelphia collar for 4 weeks and a soft collar for 1 week. Physical therapy was given for 3 weeks. The student successfully completed his training, and is flying operationally without problems.

Case 4. Left Herniated Nucleus Pulposus (HNP) C_{5-6}

Patient D was a 35 year old white male F-16 IP/Fighter Weapons School graduate, 173 cm tall, 72 kg, and of medium build. He had 1,200 hours in the F-4 and 1,000 hours in the F-16. He had been injury-free in two motor vehicle accidents, in one of which he was thrown free of the car and walked away from it. He had experienced three to four episodes of neck strain in the F-16 which he had self-treated with heat and aspirin.

Equipment: Custom fit HGU/26P helmet, MBU-5P mask.

Mission: No specific engagement.

Discussion: The patient noted a gradual onset of paresthesia in the left arm which occured only when loads greater than +6 G_z were applied. This is similar to a diagnostic observation known as Spurling's maneuver (compressing the head of a patient suspected of cervical radiculopathy; a positive result is reproducing radicular pain). He attributed these symptoms to a possible shoulder strain related to moving furniture. However, these symptoms persisted for 3 months, and he began to experience weakness in his left arm, especially when hooking up his anti-G suit. He finally presented to the flight surgeon with intermittent left shoulder pain, left biceps weakness, and numbness down his left arm into his thumb. Anti-inflammatory medication for 1 week provided no relief. He was then placed in C-spine traction for 10 days, again with no relief; a myelogram showed a left HNP at C_{5-6}. Surgery consisted of an anterior discectomy/osteophytectomy. Symptoms disappeared and he was RTFS in 5 months without sequelae. He has continued to do well in the F-16.

Case 5. Left HNP C_{5-6}

Patient E was a 29 year old white male F-16 student pilot, 168 cm tall, 75 kg, with a medium build. He had a total of 1,200 hours flying time with 700 hours OV-10, 340 Hours A-7, and 15 hours F-16 time.

Equipment: Protection incorporated light weight single visor helmet with an MBU-5P mask.

Mission: Defensive BFM (DBFM).

Discussion: The student was flying the third engagement of DBFM and had problems checking his 6 o'clock and tracking the IP's airplane under G loading. Looking over his left shoulder to 7 o'clock with maximal neck rotation he attempted to check the G-reading on the HUD (head-up display) and noted 8.4 G; he then looked back to left 7 o'clock. He experienced neck pain, which radiated into his left arm, with this maneuver. The next week he flew a repeat DBFM and experienced pain when looking over his left shoulder. He was unable to check the 4-7 o'clock position inclusive. Six more engagements of DBFM were flown and neck pain and left radiculopathy symptoms persisted postflight. The flight surgeon noted a muscle spasm in the left side of the student's neck and injected it with local anesthetic. Physical therapy and cervical traction for 3 weeks provided only slight inprovement. A myelogram showed a left HNP of C_{5-6}.

The patient underwent an anterior discectomy/osteophytectomy and was RTFS 5 months after the operation. He has done well in the F-16, but has noted occasional neck pain at 8 G and above, when his head is not properly positioned.

Case 6. Myofascial Pain Syndrome

Patient F was a 37 year old white male F-16 IP, 175 cm tall, 74 kg and of medium build. He had 3,300 hours total flying time with 1,000 hours T-38, 700 hours F-4, 720 hours OV-10, and 770 hours F-16 time.

Equipment: HGU-48P helmet (custom fit) and an MBU-12P mask.

Mission: No specific engagement.

Discussion: The patient noted transient neck soreness, aggravated by flying, for approximately 1 year. Three months prior to evaluation, symptoms began occurring more regularly, especially on 9 G student orientation rides. He would schedule to avoid these high G missions. His symptoms finally caused him to seek evaluation by the flight surgeon. The C-spine x-ray series was negative. Analgesics, and anti-inflammatory medications and osteopathic manipulation were provided. The symptoms progressed to numbness and tingling into the right medial forearm. Electromyography (EMG) findings were consistent with a right C_6 nerve root irritation. Bone scan was negative. C-spine traction was applied for 1 week. CT and myelograms were negative for HNP. The patient gradually improved after a 4-week grounding. He now denies any return of symptoms, and regularly performs warm-ups of his neck prior to BFM missions in the F-16.

Case 7. Fracture of Spinous Process of C_7

Patient G was a 36 year old white male F-16 IP, 182 cm tall, 75 kg, with a thin build. He had 2,250 hours total flying time including 850 hours F-4, 1,300 T-38, and 400 hours F-16 time. Four years prior to the current injury, he suffered a hyperextension injury to the neck secondary to a rear-end automobile collision. The injury was treated with medication for 4 days. He also had a history of two significant neck strains in the F-4, 8 and 10 years prior, secondary to "head trapping" during high G loading in the back seat as an IP. Since he began flying the F-16 3 years prior to his injury, he has regularly performed isometric neck warm-up exercises 3-4 times per week for approximately 15 minutes. There was no history of ejection.

Equipment: HGU-48P custom fit helmet/MBU-12P mask.

Mission: Offensive BFM (OBFM).

Discussion: The patient was flying as an IP in the back seat of a F-16B for a student OBFM mission. On the second engagement, the student began an abrupt 5 G climbing turn, the opposite of what the IP expected. The IP was looking over his left shoulder at the time, and felt a sharp pain at the base of his neck. The IP repositioned his head to neutral position without further pain. Two more engagements were flown (under 6 G). The IP avoided further neck positioning under G loading. Approximately 3 hours after landing he developed increasing pain. C-spine x-rays later that day showed a fracture of the spinous process of C_7 (Fig. 7.) Analgesics and a soft collar for 2 weeks, followed by gradual range of motion and strengthening exercises, resulted in improved symptoms. He was RTFS after a period of 8 weeks. He continues to perform his isometric neck conditioning exercises.

Case 8. Left HNP C_{6-7}

Patient H was 36 year old white male F-16 IP. He was 173 cm tall, 80 kg, with a stocky build. He had 4,050 hours total flying time with 1,200 hours F-16, 750 hours F-4, 1,300 hours T-38, 600 hours OV-10, and 50 hours F-5 time. At age 18 he hit the front windshield when his car hit a tree. There was no loss of consciousness or neck injury.

Equipment: Custom fit HGU-48P helmet/MBU-12P mask.

Mission: Defensive BFM.

Discussion: The patient was instructing a student on defensive BFM from the back seat of an F-16B. The first engagement involved an attacker at a high right 5 o'clock position, who then repositioned to high left 7 o'clock and proceeded to a deliberate overshoot. The student performed an unloaded 180° roll, then snapped full aft stick to 9G. At this time, the IP had been looking to the right 5 o'clock and during the attacker's transition to the left 7 o'clock the abrupt, unexpected G forced the IP's head into his lap. He used both hands to push his head back into position, and felt a burning sensation in the left midline posterior neck. He flew two more engagements and noted discomfort in his neck in the left lateral gaze position. The burning sensation persisted during the remainder of this flight, but no significant pain was noted and all symptoms resolved on landing. That evening the patient noted neck stiffness and decreased mobility, which he treated with aspirin. He flew three more times that week in the front seat up to 9 G without any significant problem. The neck stiffness gradually returned. That weekend he experienced sharp neck pain the left posterior midline which was worse when looking down and to the left. Again, he treated himself with a heating pad and extra-strength acetaminophen. The following day, while tilting his head back to shave the right side of his neck, he experienced a severe shooting pain into his left arm, which brought him to his knees. The flight surgeon noted a decreased biceps muscle mass and decreased biceps tendon reflexes on the left. C-spine x-rays showed disc narrowing at $C_{4,5,6}$. Traction initially improved symptoms; however, his symptoms gradually recurred over the next 3 weeks despite his doing only light office work. A myelogram showed a left HNP at C_{6-7}. He underwent left anterior discectomy/osteophytectomy at C_{6-7} and was RTFS after 6 months. He has had no further problems in the F-16.

DISCUSSION

Anatomy and Physiology

The cervical spine consists of 34 joints from the bottom of the skull to the undersurface of C_7. Between adjacent vertebrae there is a closed five point support system. The five points are the intervertebral disc anteriorly, two zygapophyseal joints posteriorly, and two neurocentral joints (of Luschka). Malalignment or distortion of any one of the five support elements stresses the remaining four elements. When a disc is stressed, it deteriorates. When the other elements are stressed, spurring occurs in an attempt to stabilitze. Injured ligaments also tend to heal with calcification. Chronic muscle imbalance from the initial injury may also cause stress and instability at distant levels (7). This in turn may cause spurring at sites other than the site of injury.

There also exists an important cervical locking mechanism which serves to protect the neck before vascular or nerve damage can occur. During extension, lateral flexion, and rotation of the neck, the transverse processes of the vertebra engage the top of the upper articular processes of the vertebra immediately below. Narrowing of the spinal canal is arrested by the locking mechanism, thus preventing damage to the spinal cord. The vertebral balance is maintained by the ligaments and muscular forces acting through continuous adjustment. The isolated ligamentous spine, devoid of muscles, is incapable of supporting more than 2 kg without buckling. However, it is important to note that patients with total paralysis of the cervical muscles have no clinical instability. In the clinical sense of stability, the ligaments play a primary role and the muscles a secondary role (8).

The mechanical strength of the human cervical vertebrae has been shown to be greatest between ages 20-39. At ages 40-49 the tensile strength decreases approximately 30%. (3) There also is an inverse relationship between range of motion and age: as age increases, mobility decreases.

As mentioned in the introduction, the cervical spine is capable of carrying greater loads axially than in flexion, extension, or torsion. The G-loads in the current dynamic cockpit environment are capable of causing stresses in excess of what the cervical vertebral system can safely tolerate in some individuals.

Evalution and Work-up

History is important in the evaluation. Aircrew frequently do not acknowledge the history of trauma, and the amount of force required to produce occult fractures or other injury to the cervical spine can be minimal. It is important to ascertain the direction of force as well as its mode of onset, duration, and location. Associated symptoms such as pain, weakness, sensory abnormalities (and their distributions), gait disorder, vertigo, nausea, diplopia, stiffness, or deformity should be determined. While trauma is the most likely common cause of abnormalities of the cervical spine in the fighter air-crew population, one should also consider and rule out such conditions as tumors, infections, congenital malformations, inflammatory disorders, degenerative disorders, and metabolic and vascular disease processes.(9)

In evaluating the history, it is important to differentiate non-neurological lesions (myofasical) from other types of cervical spine injuries. These extraspinal injuries are often called "burners" because of the searing pain they produce in a radicular distribution (10,11). To assume that the symptoms are produced by a herniated disc is potentially to mismanage the patient and possibly ground him for unnecessarily long periods of time.

Occult lesions also must be considered. The usual sequence of events is neck pain immediately following injury. The pain then becomes less intense or even disappears as the muscles of the neck go into spasm and act as an internal splint. The aircrew member may not report for evaluation at this time. However, as the muscles tire, the pain recurs or worsens. Symptoms may persist, or they may disappear and perhaps return a third time. An occult injury should be considered with such a history of recurrent neck pain, even if the patient cannot recall any precipitating trauma (7).

Physical Examination

The physical examination should attempt to separate musculoskeletal from neurologic injury, to elicit instability, and to distinguish upper motor neuron from lower motor neuron signs. It is important for the examining physician to be familiar with the entire spectrum of findings associated with cervical spine injuries.

Diagnostic Tests

C-spine x-rays remain the initial evaluation tool for suspected C-spine injuries. In acute injury situations, where C-spine injury is suspected, a lateral scout film to include C_7-T_1 should be obtained. The scout film is reviewed for evidence of fracture, dislocation, or instability before anterior-posterior, lateral flexion/extension, and oblique films are obtained. Occult traumatic lesions are usually not appreciated on the first set of routine C-spine reviews and may require special views in numerous projections for their detection (9).

Myelography, CT, or MRI may also assist in further evaluation of spinal lesions. Bone scintigraphy (scan) with ^{99}Tc-methylene disphosphate (MDP) or ^{99}Tc-hydroxymethylene diphosphate (HDP) are useful in detecting active disease (osteoarthritis) and acute injuries (compression or stress fractures). MDP can assist in dating injuries, as bone scans of the majority of spinal fractures return to normal in 6 months (9). Electromyography can also help distinguish between spinal cord, brachial plexus, and peripheral nerve lesions when other studies are inconclusive. (Further information on the above tests and current advancements may be found in the current orthopedic, neurosurgical, and radiologic literature.)

Treatment

It is beyond the scope of this paper to discuss in detail the treatment and management of the various combinations of cervical spine injuries that can occur. Appropriate orthopedic or neurosurgical consultation should be obtained when specific injuries are suspected or identified.

Cervical collars remain a popular modality of treatment for a wide variety of neck injuries treated on an outpatient basis. An understanding of their limitations is important in managing patients with neck injuries. If the stabilizing ligaments have lost their integrity, a halovest or halocast may be required. Cervico-thoracic braces and braces with mandibular support limit a high percentage of motion in the sagittal plane, but they are less effective than the halo in restricting rotary or lateral motion (18). Soft collars provide gentle support and act as a reminder to the patient to limit neck motion: they do not immobilize. The collars should be worn as long as it takes for the fracture, ligaments, or soft tissue to heal. Generally, the collars are worn for a minimum of eight to ten weeks, depending on the type of injury. It is better to err toward longer rather than shorter wear. This is frequently a difficult task with most fighter aircrew.

It is important that the stability of the neck be assessed after healing and before the crewmember returns to the high-G environment. In addition, he should be strongly encouraged to undertake neck strengthening and conditioning before resuming full flying activity.

PREVENTION

Physical Conditioning

There are a variety of neck strengthening exercises, including neck bridging, buddy system resistance exercises, the use of a head harness and/or straps with free weights, and neck resistance machines. Many of the exercises are awkward, lack specificity, and produce less than optimal results.

A number of neck conditioning machines have been developed for protecting football players' necks. Studies done with West Point cadets have shown significant increases in neck strength using such machines (12). These machines provide resistance through the full range of motion and do it safely. The described workout consists of six exercises (all six of which took 8 min): shoulder shrug,

neck rotation, and exercises on the 4-way neck machine (flexion, extension, right and left lateral flexion). In one study, one group performed whole body conditioning along with neck training twice a week for 6 weeks; a second group performed only neck exercises three times a week; and a third (control) group was included with no formal neck training program. The increase in relative strength of the neck noted was 92%, 57%, and 28%, respectively. This study indicates that: 1) total body condidtioning is important in achieving a significant increase in neck strength, and 2) a brief neck-exercise training program (8 min, two times a week) can give a significant increase in neck strength when performed on proper equipment. Many fighter aircrew, especially those in the F-16 community, have developed some form of neck conditioning on an individual basis. Various harnesses with free weights have been designed by aircrew for exercising their necks and have been recommended in the flying safety literature (13). Most fighter aircrews perform neck warmups in the cockpit prior to taxiing or while waiting for takeoff. This conditioning becomes even more important for the older crew-member and the infrequent flier.

Head Positioning in the Cockpit

Various techniques have evolved for head positioning in the high-G environment. In the F-16, some aircrew position their head prior to the high-G onset, and if repositioning is required, the aircraft is unloaded (G load is decreased) and the head is repositioned. Others "wedge" their head against the edge of the seat or canopy to check the six o'clock position while under high G loads. Still others are able to move their head around in the 9 G environment without apparent difficulty. It is also interesting to note that many pilots (especially older ones) prefer, and find it easier, to look over their left shoulder rather than the right, and will often arrange defensive engagements to meet this preference. This may be due to the traditional placement of the stick between the legs and a left hand throttle making it easier to look left than right. This does not seem to be a habit in new F-16 pilots.

In the F-15, which has the same ejection seat as the F-16, the entire spine becomes involved in support because of the relatively vertical positioning. Aircrew will often push-off on the canopy with one hand to brace themselves, or use one of the "towel racks" (canopy handholds) to assist in their viewing (Fig. 8). Because of the more vertical configuration of the seat, it is possible for the head to fall further forward if the momentum of the head pulls the torso forward (Figs. 9, 10). It is also more difficult for the pilot to visually scan vertically (directly above the canopy) (Fig. 11). Physical strength becomes more important with this seat position. Figures 8-11 show various viewing positions in both the F-15 and F-16.

Equipment

Helmets have continued to be improved since the P-1 helmet was adopted in 1948. Current single-visored light-weight helmets, such as the HGU-26/P (approximately 1 kg) reportedly are more comfortable and cause less fatigue under repeated G loading than was the case with the older helmets. Protection of the neck during ejection and impact continues to receive research attention. This research includes the development of mathematical models to evaluate the biodynamic response of the head and neck to ejection and impact forces (8).

The basic premise for neck protection is to avoid distortion of the neck beyond its mechanical limits. The most direct means of preventing injurious neck distortion is external reinforcement of the cervical spine, head to torso. However, a device capable of such reinforcement must also be comfortable to wear without being a hazard or causing interference during normal or emergency tasks. Several interesting designs were considered in the late 1960's by Mattingly (14)(Fig. 12). Other concepts of protection are seen in articulating seats which can provide head and neck support in multiple viewing positions under varying G loads as proposed by McDonald (15,16) (Fig. 13). Further work in the area of neck protection equipment remains to be done as aircraft continue to improve in performance capability.

Screening of Aircrew

Recommendations for the medical screening of potential fighter aircrews, as well as for recurrent examinations of current aircrews, have been proposed (17,18,19). The Royal Netherlands Air Force (RNLAF) instituted medical screening in December 1982, with their introduction of the F-16 (20). This screening consists of 10 x-rays: 4 lumbosacral spine (PA/lateral/2 obliques), 2 thoracic spine (AP/lateral), and 4 cervical spine (PA/lateral at rest and in flexion and extension). This gives a calculated exposure dose of 650 mrem. As a result of this examination, 20% (45/225) of aircrew applicants were rejected for x-ray evidence of spinal abnormalities. Only 2% (4/225) were disqualified because of cervical spine abnormalities. The RNLAF also examined 196 qualified fighter pilots and found 24% (48/196) with cervical disorders. Four pilots were rejected from F-16 duty because of cervical discopathies with osteophyte formation (especially when the osteophyte involved the back side of the cervical canal). Three pilots received a G restriction.

The French Air Force has adopted a similar program (21). However, statistics from this program have yet to be published. Presently, the USAF does not require spinal x-rays for fighter aircrew candidates. It is doubtful that any of the eight cases presented in this report would have been prevented by initial or recurrent screening.

CONCLUSION

The potiential for significant neck injury exists within the current operational envelope of today's high performance fighter (HPF) aircraft. Although the known cases of significant neck injury due to high G stress are few, and have not resulted in permanent grounding or disability, the number of tactical aircrews currently on flying status and the number of sorties flown on a daily basis suggest that the possiblity exists for more serious or catastrophic neck injury. Aircrew experience level is no protection against neck injury, and G-load exposure need not be excessive to produce injury.

Exercise conditioning of the neck has significantly reduced the frequency of neck injuries in football players and can offer significant protection to today's fighter aircrew. Total body conditioning is preferable to simple neck exercises. A variety of methods and types of equipment are currently available to achieve these goals, some better than others. Twice a week conditioning of the neck can give significant increases in strength over a short period of time. Finances, personal preference, and available space are all factors that have led to a variety of conditioning programs by aircrew. Aircrew members should be instructed in safe methods of conditioning, and should also be taught how to recognize significant symptoms or injuries.

Equipment (helmet and mask design) has improved, but seat designs and other means of neck support deserve further research, especially for the next generation of fighter aircraft.

Screening of USAF HPF aircrew candidates, as well as present fighter aircrew, with spinal x-rays is currently a sensitive issue among both aircrew and the medical community, and whether such screening will eventually become standard practice is unknown. Nevertheless, flight surgeons who support HPF aircrew must maintain a high degree of awareness of the possibility of a serious neck injury when caring for aircrew who present with neck complaints.

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SUMMARY OF NECK INJURIES

Case No.	Injury	Mechanism	G Load($+G_z$)	Mission*	Aircraft
1	Compression Fx C_7	Acute flexion	9 G	NBFM	F-15A
2	Compression Fx C_5	Forced lateral gaze	6.5 G	DBFM	F-15A
3	Interspinous ligament tear at C_{6-7}	Strained on right posterior gaze	4.5-5.5 G	DBFM	F-15A
4	Left HNP C_{5-6}	No specific injury	Sx at 6 G	--	F-15A
5	Left HNP C_{5-6}	Strained on left posterior gaze	8.4 G	DBFM	F-16A
6	Right C_6 myofascial syndrome	No specific injury	Sx at 9 G	--	F-16A/B
7	Fx spinous process C_7	Neck trapped in left gaze with unexpected right turn	5 G	OBFM	F-16B
8	Left HNP C_{6-7}	Neck trapped during transition from right gaze to left gaze	9 G	DBFM	F-16B

*Mission:
- NBFM - neutral basic fighter maneuvers.
- DBFM - defensive basic fighter maneuvers.
- OBFM - offensive basic fighter maneuvers.

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The opinions, interpretations, and conclusions contained herein are those of the author only and do not necessarily represent the official views, policies, or endorsement of the USAF, Air National Guard, or any other governmental agency. Major portions of this paper have been published in "Aviation, Space & Environmental Medicine" 1989.

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FIGURES

Fig.1. Case 1-Position of neck before and after collapse of neck muscles.

Fig.2. Case 1-(left photo)-Lateral cervical spine X-ray showing 22% compression of C_7.
Fig.3. Case 1 (above)-Bone scan 7 days post injury showing uptake at C_7.

Fig.4.-Case 2-Lateral cervical spine X-ray showing a compression fracture of C_5.

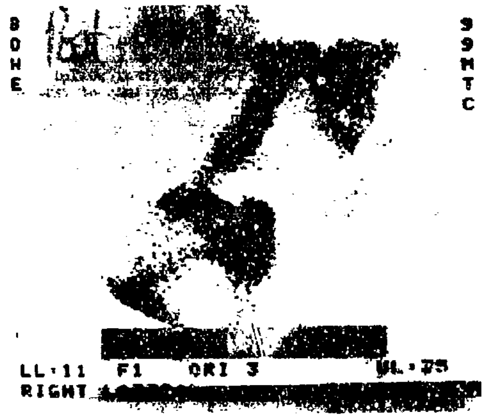


Fig.5.-Case 2-Bone scan performed post-injury showing uptake at the C_5 level.

Fig.6.-Case 3-Lateral C-spine showing widening of the C_{6-7} interspinous ligament.

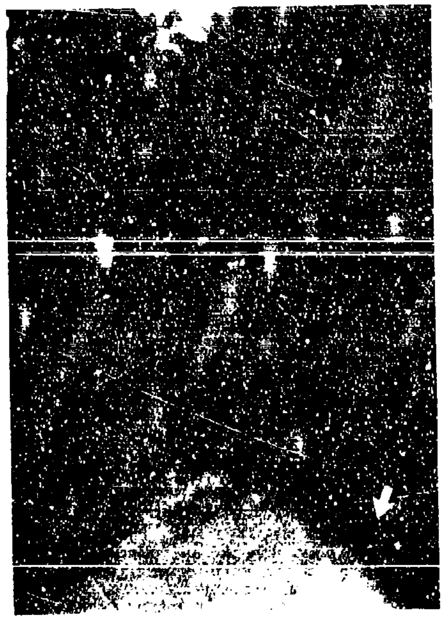


Fig.7.-Case 7-Cervical spine X-ray demonstrating a clay shoveler's fracture of the spinous process of C_7.

Fig.8. "Checking 6" in the F-16, demonstrating how the head must be positioned off the seat to view.

Fig.9. Neck at normal repose in the cockpit of the F-16(left) and the F-15(right).

Fig.10. Forward flexion in the F-16(left) and the F-15(right). Note that in the F-16 with the 30° seatback angle, the cervical spine is already flexed, thus decreasing the distance it must travel to reach maximal flexion. In the F-15, the seat is relatively straight and the cervical spine relative to the thoracic spine, is capable of further motion/acceleration relative to it's starting position.

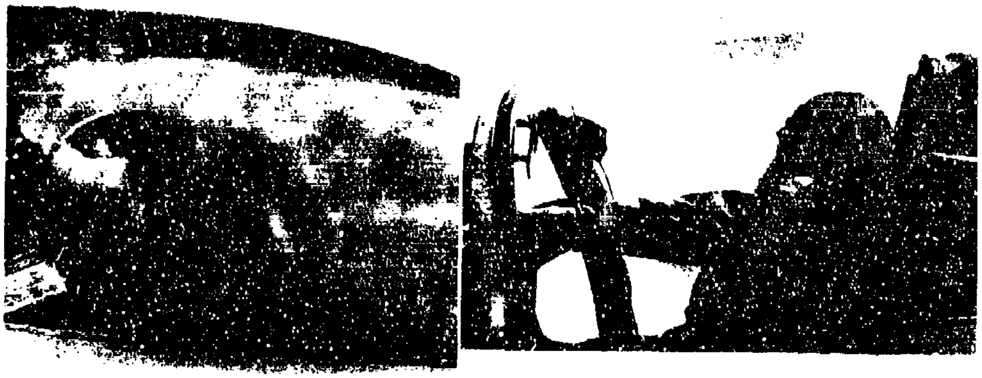


Fig.11. Viewing the vertical in the F-16(left) and the F-15(right). Note in the F-16 the torso against the seat with head back against headrest. In the F-15, note how the pilot rotates his torso, pushing off with his left arm on the hand hold and his head back against the canopy.

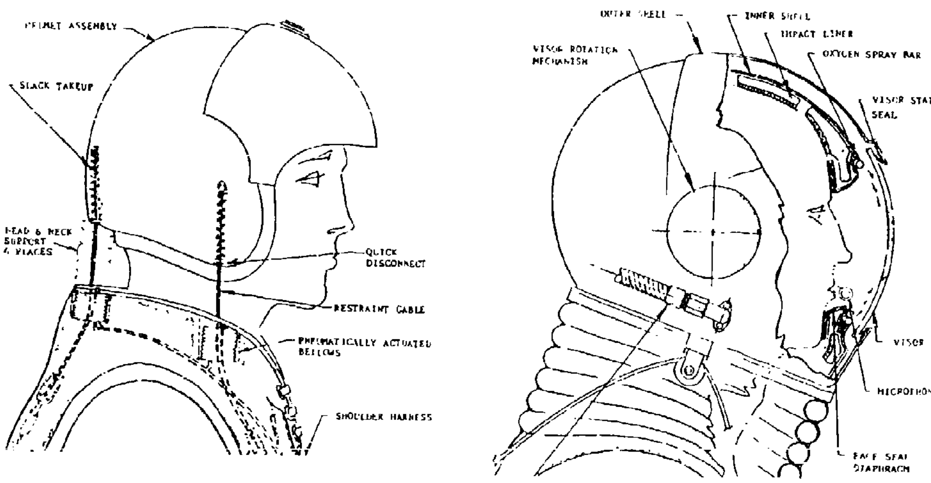
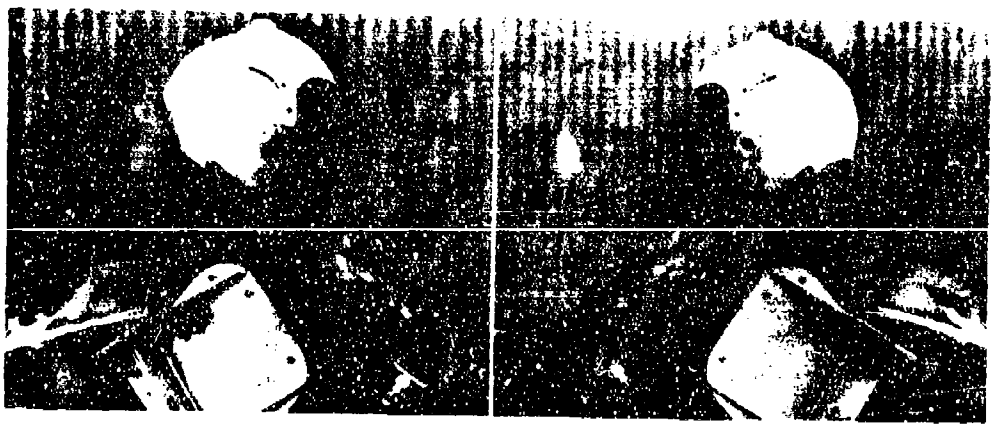


Fig.12. Concepts for Neck protection as proposed by Mattingly(14).
Fig.13.(Below) Articulating seat/headrest as proposed by McDonald(15,16).



A SURVEY OF CERVICAL PAIN IN PILOTS OF A BELGIAN F.16 AIR DEFENCE WING.

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SUMMARY

Since the F-16 replaced the F-104 Starfighter in 1977, the pilots ot the 1° Fighter Wing (1°FW) complained frequently from neck injuries sustained during and after their High + Gz interceptions. Till recently, Aerospace Medical community paied few attention to this new clinical problem, although it was wel known amongst pilots flying high performance aircraft (HPA) and flight surgeons.

In this report, we communicate the results of an anonymous questionnaire, concerning neck problems in pilots flying the F-16 in an almost exclusive air to air role. A sample of 30 pilots answered this questionnaire in 1984 and in 1988.

Analysis of these questionnaires showed 50 percent of our pilots reported having neck problems flying F-16.
No positive correlation between the age of our pilots and the prevalence of cervical pain could be demonstrated in this small group of fighter pilots. Subsequent factors favourising thes neck injuries, are the weight of the flying helmet as well as the combination of an inappropriate and insufficient physical training program.

Neck injuries in pilots of HPA are a real occupational hazard and we need further long term follow up studies to assess an eventually cumulative effect of repetitive high G loading on the cervical spine.

INTRODUCTION

In the beginning of the seventies, a new generation of high performance aircraft (HPA) was developed by the aeronautical industry. The application of new technologies, microminiaturization, the fly by wire system, the use of strong light weight composite materials and new powerful engines, led to the development of a new light weight fighter (LWF) : a highly maneuverable aircraft, combining rapid linear acceleration with a high onset rate of centripetal radial acceleration (+Gz). For the first time in aviation history, the pilot has become the limiting factor in the man-machine combination.

In the 1st Wing, stationed at Beauvechain AB, BE, the first F-16 was delivered in early 1979 to replace the F-104 Starfighter. On the first of january, 1981, we had the first operational European F-16 Squadron assigned to NATO Airforces. At present, 168 pilots have received a conversion course at the Operational Conversion Unit Squadron (OCU) with 70 percent of their missions flown in a high G environment.
Early in 1980, flight surgeons at the 1° FW had already noticed a very high incidence of cervical injuries among fighter pilots, once they began operational flying in their assigned air defence role. In an anonymous questionnaire, 22 out of 23 interrogated pilots complained of cervicalgias, an unknown medical problem in the fighter community at that time. We recommended strongly that new flying helmets be purchased to reduce the high-G load on the cervical spine.
We advised F-16 pilots to perform a general muscle training program, with special attention on the development of the neck muscles, and we proposed that the Belgian Air Staf fund a kinesiotherapeutic treatment facility on the base for treating the pilot's neck injuries.

Although many HPA pilots over the world have suffered an acute inflight neck injury, and flight surgeons were aware of the potential problems, there was little published material until recently. In 1985 a radiologist from the Royal Netherlands Air Force reported the results of a radiological investigation of the whole spine with special interest on the cervical spine. For pilot training candidates, there was a rejection rate of 20 percent for spinal radiological disorders. Of 196 qualified pilots radiologically examined, 18 showed cervical discopathies with osteofyte formation. After deliberation, four pilots were rejected for F-16 duties and two others received a G-restriction limitation.

The medical Service of the Belgian Air Force decided in 1984 to take cervical spine X-rays of all current F-16 pilots. Although no qualified pilot was grounded for radiological disorders, the BAF Medical Service, decided to repeat this examination at five year intervals as follow-up study.

MECHANISM OF AN INJURY.

Pilots flying air defence interception missions are particularly at risk for neck injuries. Dissimilar Air Combat Training (DACT) missions, flight safety regulations and rules of engagement require a near constant visual contact of the opponent aircraft. Particularly when in a defensive position, the pilot must rotate his neck to check his six o'clock position with heavy load on his aircraft.

The cervical spine, compared to the dorsolumbal spine, is far more mobile, but this high degree of mobility is penalized by its greater fragility. The anatomical structure of the cervical spine allows a person to make complex and large movements with his head (flexion, extension, lateral flexion, rotation and a combination of these). Specialised articulations between the neck vertebras permit this complex mobility with minimal muscular activity, while strong licaments, anterior, posterior and interspinous, under normal conditions limit excessive mobility of the vertebras. However, if these supports are damaged, larger displacement of the vertebras becomes possible and can allow damage to the spinal nerve roots.

The neck muscles form a kind of active rigging of the head and the cervical spine. Since the center of gravity of the head is situated before the atlanto-occipital articulation, the head has a natural tendency to fall forward which must be countered by a constant contraction of the strong anti-gravity neck extensors. On the other hand, the neck flexors muscles are small and weak. A pilot expecting high G forces, can contract his neck muscles and stabilize his head in a steady position. If, on the contrary, the pilot is caught by a surprising high G load and his head is not stabilized, the head will move very rapidly in the opposite direction of the applied acceleration. In these circumstances, the neck muscles cannot react in a timely manner to stop the large displacement of the head. Further they lack sufficient strenght to return the heavy weight of the G loaded head to a stable position. The displacement of the head will only be halted by the anatomical structures of ligaments or vertebras. Such a chain of events can typically lead to a muscular elongation, wich can trigger a pain-spasm reflex leading to a torticolli; a ligament tear or in the worst cases (sub) luxations, intervertebral disk damages, compression fracture of the cervical vertebras and possibly compression of the nerve roots from the plexus cervicobrachialis.

It is quite obvious that the weight of a heavy flying helmet of about 2 kg will considerably increase the load on the cervical spine. Yet another deleterious effect of the older flying helmets is the synergetic combination of the double visors and the oxygen mask fitting placement, which shift forward the center of gravity of the head increasing the load for the neck extensor muscles.

METHODS.

In 1987, the authors functioning as flight surgeons at the 1° Fighter Wing, had very few medical consultations with pilots suffering with neck problems, although it was known that these complaints were very common in 1980. When we saw a pilot with neck problems, it was typically a trainee at the OCU Squadron undergoing training in Basic Fighting Manuevering (BFM) or a young squadron pilot with limited F-16 flying experience. Flying in the back-seats F-16 B Models during BFM missions, we experienced ourselves the heavyload on the neck muscles and questioned why we saw relatively few pilots with neck injuries. The answer to this question could be two-fold : the problem wasn't as significant as 8 years ago, or the pilots, afraid to see a flight surgeon for fear of being grounded, preferred to seek medical attention for their neck problems with a civilian physician. Informal talks with the pilots, confirmed the continuing existence of neck pain during air combat maneuvering. In these discussions, most pilots attributed this problem to their heavy, bulky, and uncomfortable flying helmets.

Thirty pilots of the 1° Fighter Wing were screened for the prevalence of cervical pain in 1984, utilising an anonymous questionnaire during their annual medical check-up at the Center of Aerospace Medecine. Since the results of that survey were never analysed, in 1988 the BAF Director of the Aeromedical Services tasked the medical detachement of the 1° F.W. to repeat the same 1984 investigation and complete a formal analysis of both data sets.

These questionnaires have been answered by 30 pilots of the 1° Fighter Wing, chosen at random. All respondents were operational pilots from the two squadrons or instructor pilots at the OCU squadron. Because there is a relatively high turn-over of pilots in the Wing, the 1984 sample group was not identical to the 1988 group. However, they were similar in the most important way, all did the same job : air defence missions.

In the first part of this questionnaire we surveyed introductory parameters such as age, weight, height and total F-16 flying hours. Later sections searched for causal factors and frequency of neck injuries. We also tried to determine the different therapeutic possibilities chosen by the pilots as well as the persistence and frequency of occurence of their neck complaints. We also investigated the existence of a physical training program and prefered sports activities. The comfort level of the present flying helmet and the position of the pilot during high +Gz maneuvers were surveyed.

RESULTS

1. General information.

The thirty respondent pilots from the 1° FW averaged 450 total F-16 hours in 1984 and 735 hours in 1988. In a normal flying day, they flew one mission with a

duration of approximatively one hour. Exercice tasking increased the operational sortie level to 2 per day, and OCU IPs occasionally flew 3 missions per day. In 1984 the pilots reported reaching an average high +Gz peak level of + 7 Gz, five times during a mission with 2 periods of high sustained Gz for more than 15 seconds. In the 1988 survey, the average high + Gz peak level decreased to + 6.4 Gz, but with 6 occurrences during a mission and more than 3 periods of high sustained + Gz.

The average age from the sample of 1984 was 30.5 years with a minimum of 22 and a maximum of 45 years. In 1988 the average age was 32 years with a minimum of 22 and a maximum of 44 years. Subdividing the samplegroup into 3 age subgroups, we have in 1984 13 pilots in the group 20 - 29 years, 11 between 30 and 35, and 6 pilots more then 35 years old. In 1988 12 pilots fell in the first group, 9 in the second group, and 9 in the third group. The yearly flying time was very low in 1984, averaging 125 hours for a squadron pilot to 150 hours for an OCU IP. In 1988 the squadron pilots averaged 170 hours while the OCU IPs flew more than 250 hours a year.

2. First incidence of neck pain.
In 1984 13 out of 30 pilots reported cervicalgias incidents since beginning flying the F-16 in the 1 FW. In six cases the first incident had a sudden onset and 2 pilots noticed an irradiation of the pain. Two other pilots described the pain as incapacitating and stated that they were forced to abort their missions, feeling that the pain jeopardized flight safety. Three other pilots reported continuing their mission, but with a limitation in aircraft maneuvering, while 7 others flew the scheduled mission with only minor distraction from their neck pain. The pain persisted in most cases from 2 to 4 days but only 50 percent of the pilots sought medical attention. These pilots were grounded for an average of 4 days with a treatment consisting of rest, kinesietherapy and occasionally a myorelaxant and/or an anti-inflammatory drug prescription. The average age of the pilots with cervicalgia was 31.5 years; the average age of those without neck pain was 29.7 years. Statistical calculation in this small sample group and between the age subgroups showed this to be of no statistical significance.

In 1988, 16 pilots reported having neck problems in flight. One pilot suffered his first incident performing as a student pilot his initial training. The 15 others reported their first neck injury flying the F-16. As in 1984 50 percent of the cases involved, noticed a sudden onset of the pain, however there were no reports of an irradiation of the pain. The injury persistence was about the same as in 1984. Two pilots also reported that the intensity of the pain was so incapacitating that they had to abort their mission. Only 50 percent consulted the flight surgeon and these were treated in a similar manner to 1984. Statistical analysis of the pain group and the non injury group average ages (33 and 31 respectively) revealed no statistical significance.

3. Subsequent injuries
About 50 percent of the pilots surveyed in 1984 and 1988 complained of injuries in flight every month. 20 percent reported weekly injuries and 1 pilot complained of encountering neck problems each flight. These cervicalgias were generally associated with a 7 G loading and not only interfered with the pilot's concentration during his mission but also with his off duty life as well. Here also, preferred medical treatment was physiotherapy although some pilots admitted seeking manual therapy.
The intensity of the pain was in most cases not as severe as the first injury, and the pain usually exceeded after a night's rest.

4. Body position during high + Gz maneuvering.
With few exceptions, the majority of the pilots reported never using the headrest of the ACES II ejection seat during combat maneuvering. As a technique to protect their neck, some pilots positioned their heads just prior to a high - G turn and changed the position of their head only after unloading the aircraft. Other pilots supported their head with the left hand, the left arm pinned with the elbow on the left "towel rack". Others wedge their head during high G maneuvers in the space between the canopy and the edge of the headrest of the ejection seat, facilitating the pilot's check of his six o'clock position. Another technique consisted of leaning a bit forward and starting to turn the torso from the waist. This increased the six o'clock coverage as well, and was sometimes aided by pulling with the left hand on the right "towel rack".

5. Flying helmet.
In 1988, a substantial number of our pilots still wore the flying helmet HGU/2 AP, a custom fitted HGU/26P. The total weight of the helmet, oxygen mask and double visor assembly included, is between 2 kg and 2.2 kg varying directly with the helmet size. In both questionnaires, this flying helmet was generally condemned by the pilots. Besides the heavy weight, the upwards vision is restricted under heavy G load due to the helmet's tendency to slip forward. This particular flying helmet has been recocnized as a serious defiency for a decade by pilots in the 1° FW. A new light weight helmet is scheduled to be delivered during the first half of 1989.

6. Pilots and physical fitness.
In the 1984's survey only 20 out of 30 pilots reported regular participation in sports activities. Favourite sports were tennis, swimming, walking, jogging and squash and to

a lesser degree cycling, soccer, powertraining and volley-ball. In 1988 only 50 percent of our pilots had a regular sports program. Part of this decrease can be explained by an increase in the above 35 age group compared to 1984 (9 versus 3), but even in the younger group between 22 and 29: 4 out of the 12 surveyed pilots reported having no regular physical conditioning program. Participation in a supervised sports activity has not been mandatory. However, each squadron has its own NCO sports instructor available, who can advice and assist the pilots during sports activities.

DISCUSSION

Neck injury, in this particular sample group of fighter pilots in an F-16 air defence wing, has been a very common occurrence. 50 percent of sampled pilots reported neck pain in flight and 70 percent from these pilots encountered this problem on a regular basis. Within this sample of 30 pilots we could not find a statistically signigicant positive correlation between the age of the pilots and the appearance of neck problems, a finding backed by the personal observations of the flight surgeons.

The majority of the pilots treated for neck injuries have been young trainees in the OCU squadron, i.e. pilots with a limited experience in ACM. Older pilots with more experience in ACM-ering have developed a higher situational awareness that allows them to predict, and counter, their opponents maneuvering. These earlier reactions can often be done at lower G loading than a maneuver initiated at a later time during the fight sequence. Further by trial and error, most of the experienced pilots have developed a protective technique for their neck during a dog fight.
However, in a fight between two equally skilled F-16 pilots, the fight will generally be won by the pilot with the best physical conditioning; the one best able to sustain the high G loading without G-loc and with strong developed neck muscles necessary to keep his target in sight during high G maneuvering. The data analysis also revealed that cervicalgias in a HPA, the BAF F-16, after ten years experience in an air defence role, are not as frequent as in the early years. This is due to many factors and we would caution the pilot population that wearing a light flying helmet is not a panacea. Although it significantly reduces the load factor for the cervical spine, it will not alone provide security from serious neck injuries. A conscientiously followed neck exercising program is the best guarantee for avoiding serious neck injuries in HPA and making such a program mandatory would undoubted ameliorate this situation.

Worth special emphasis, is the lack of a single reported case of serious structural injury to the spinal column. In ten years of F-16 flying, involving almost 55.000 flying hours, only two cases of serious musculi-ligamentair neck injuries were encoutered that needed more than 4 weeks to recover. Our fight population has been spared the injuries, like herniated nucleus pulposus or cervical compression fractures, diagnosed in other countries flying HPA like the F-16.

Neck injuries would appear to be a real occupational hazard for fighters pilots flying HPA. Long term comprehensive studies, assessing the possible cumulative effects of repetitive high G loads on the cervical spine during a career of 10 or 20 years in the F-16 are in order. Towards that end, the BAF Medical Service started in 1984 taking cervical spine X-rays of all pilot candidates. The aim is to repeat these X-rays every 5 years and compare the results of F-16 pilots with a control group of pilots not flying the F-16. The target survey study will be extended in the future to the tactical F-16 Wings, although, having a primary air to ground role, they perform air to air missions on limited basis.

A futur development warranting concern, is the proposed use of the helmet mounted display whose additional weight could offset any gains received by the light weight helmets currently being purchased. Without proper attention from the Aerospace Medical community, Dr. Vanderbeck's statement, foreseeing the cervical spine as the high G "Weak link" of the human system could well become reality.

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DISCUSSION PERIOD 1

Van den Biggelaar, Netherlands

I have a question for Dr Vanderbeek about his interesting presentation. His figures are made up from three sections: the F5 population, the F15 and the F16. Did you include the pilots flying in the high G aggressor role in your F5 population? That is the F5E I believe.

Major Vanderbeek, USA

Yes, the operational F5 wing was the aggressor squadron.

Dr Von Gierke, USA

What percentage of your pilots can relate their injury to a very specific event or G manoeuvre, and what percentage had it only after long exposure to the flying environment?

Major Vanderbeek, USA

I did not ask them to relate their injury to a specific in flight occurrence so I cannot answer that. If you ask them what manoeuvres or neck positioning results in injury they would almost all say either checking 6 o'clock (or 4 or 5 o'clock) or moving their head under high G loading. Most pilots will have a specific head movement that they will not do ever again because they know that spot will cause a reinjury similar to one in the past.

Col Hickman, USA

I had one or two questions I would like to ask all three of the last participants. I was struck by the fact that after an acute injury almost everyone continued to fly, both in that engagement and with multiple following engagements. In addition they subsequently flew other missions. How many of you are aware, as operational flight surgeons, how often an aircrew member decides to knock it off during the mission and stop flying? It must be extremely rare.

Answer

Yes it is very rare. I have only known of one individual pilot who called 'knock it off' at the time of his injury, most will defer that information until they get down on the ground or a few days later when they may relate it to their flight 'mates'.

Answer

We have a love/hate relationship in the flight surgeons office and I think a lot depends on the confidence level the pilots have in the flight surgeon. I think another important component is the importance of accomplishing the mission and filling the training slots. The weather is not a big factor in Arizona but the curriculum is run at a high pace, so the instructors are reluctant to abort a mission because their neck is is injured. They will not come and tell you unless you are hanging down at the squadron, and you notice the guys are walking around and moving their necks and shoulders as a unit. Then you approach them, but that is generally your only clue.

Van den Biggelaar, Netherlands

Way back in 1982/3 the F16 Medical Working Group was founded by colleagues in the NATO nations flying the F16. Neck problems were discussed in this group many times and were even challenged in official papers. As I recall it in one paper in 1985 the pilot population was not prepared to discuss neck problems at all; they would much rather not go to the flight surgeon in order not to be washed out from flying. But there was a problem, right from the beginning.

Captain Brooks, Canada

I am curious to know if you did a control study and took a look at a matched group of, say, groundcrew to see if sports or anything else was possibly making this a bigger problem?

Answer

No I did not.

Col Asleben, GAF

Col Schall you were showing some of your slides of the F15 showing that they have grips to stabilise the position of the pilot; on the other hand they might mislead into the wrong position. In earlier aircraft, the F4 for instance, the pilot had a better position when he flies and turns and pulls G. I know the G onset rate is not as high, but have you ever discussed the operational requirement versus the medical requirement to have a proper position and not using these grips to position yourself in the wrong place whilst you are pulling high G

Col Schall, USA

To specifically answer your question. No we have not discussed proper positioning in the cockpit. Most of the pilots learn this on their own while they go through what we call RTU training, which is the training they get in the particular aircraft or weapons system they are being trained in. In the examples that I showed, not all pilots use the handle. Many pilots will position their hand onto the plexiglass of the canopy and use that to brace themselves to look out. So if it happens to be convenient to use the hand holds they will but not everyone does that.

Van den Biggelaar, Netherlands

The problem with the pilot is that he does not really care about his neck, he cares about his enemy. He wants to see his enemy before he shoots him. In other types of aircraft to the F16 you don't really need to brace yourself.

Answer

I might just add that it is becoming more popular to acquire the target, turn your head back and forward, apply the G as desired, release the G, turn your head back and reacquire the target, thats becoming more accepted as a preventive measure.

Dr Von Gierke, USA

I would like to ask an unfair question to all three speakers and perhaps to Wg Cdr Anton. Have any of these neck injuries been implicated in accidents, and as we heard at other AGARD meetings, checking 6 has been implicated with respect to G-LOC. One is not quite sure if it adds to it or not, but I guess some people suspected it. Now could it be that some had some acute cervical trauma that led to an accident.

Speaker.

None that I am aware of.

Speaker.

I am not aware of any specific engagement in which cervical spine trauma has been listed as a primary cause of an accident. However in the first case that I presented had that pilot been flying by himself he would have been destroyed and we probably never would have known the cause of his accident.

Wg Cdr Anton, UK

I am not aware of any cases that have lead to an accident either. But I think it is of note that the three cases we have had in the Royal Air Force where people who have hand in-flight neck fractures, have always been the non-flying pilot who has been caught unawares The circumstances of the flight are such that you wouldn't expect it necessarily to lead to an accident.

Col Hickman, USA

It is clear that the problem is very under reported and therefore not very well treated. But suppose that with a major educational effort a higher percentage of thses injuries were reported to you . No 1 What is your threshold for removing someone from the cockpit for a period of time. Suppose that we had better reporting, how often would you say we would need to remove someone. If all of the major injuries needed to be grounded for a while, which would be about 9% in three months, and if only 25% of all the rest needed to be grounded we would be removing from the cockpit maybe 20% of all the fighter pilots every three months for a period of time. No 1 What is your threshold for removing people from the cockpit and No 2 how often do you think we would have to do it and what do you think the operational impact would be.

Speaker

I can only speak for myself. My criterion for removing a pilot is based on his range of motions and I have them demonstrate that to me at the office. Obviously this is not under G loading, but one frequently gets a chance to fly with these people if there is any question about their neck. At the base I was stationed at we had quite a large number of two seat aircraft available so scheduling a two seat aircraft to do this was not a problem, although I did not have to do that. I think the approach to education is a two prong approach. Firstly you need to educate the aircrew and secondly you need to educate the flight surgeons that are taking care of them. I think that most of the flight surgeons I came into contact with, did not have an appreciation for the loading the cervical spine can tolerate in various positions and sustain injury. This certainly generates other areas of controversy as to the role of cervical spine screening and we as flight surgeons try to be advocates for the pilots and I try to tread very carefully we don't necessarily want to subject a lot of pilots to myelograms and other types of studies that are invasive; and yet we want to be an advocate for them. if they have an injury we want to protect them.

Speaker

I haven't thought about that question a lot as far as a specific threshold is concerned. I think it would have to be developed individually with each pilot, based on his functional capability, and I have not really thought about how I would determine whether or not it was safe to fly or not. Once he demonstrated functional capability I would probably recommend a non-demanding sortie for a couple of rides to make sure that the assessment was in fact realistic.

Col Hickman, USA

Yes I think that is really important if we are going to say that we have an epidemic it ought to be based on things we would ground people for.

Wg Cdr Anton, UK

I think Dr Hickman's point is an interesting one. One of the things we see with our aircrew flying Hawks, very few of whom will go anywhere near the doctor with neck injury, is that they reschedule their program amongst themselves, and so if an instructor has got a relatively painful neck that seriously limits his flying his colleague will shift into his slot for a day or two and he will go and do another job in the squadron. What we don't know, and what is I think perhaps the really interesting question, is if you take an exercise and fly people in combat sorties day after day, how many people are going to be generated by day 2 or 3. I think that is the question that still remains to be answered.

Speaker.

I would like to comment having just flown in these kind of exercises and taken care of an F15 crew, deployed in that type of environment. My personal observation is that it was not a problem although they generated quite a few sorties a day for each of the different days of the war. In various types of environment both air to air and air to ground it did not seem to be a problem.

Speaker.

I would just like to add to that that at my base we took two weeks and flew nothing but clean F16's and flew pure air to air BFM manoeuvres. After the first 3-4 days there were several pilots who were glad that their 4 day experience had ended so they could rest for the weekend to prepare their necks for the second week of the exercise. What the impact would be in a real combat I'm not sure, but it is somewhat cumulative over time for repetitive close in flights.

Professor Snijders, Netherlands

I have a question for Dr Biesemans, it concerns his statement when he emphasised the importance of exercise fitness. In general I agree that fitness would be a good protection against this sort of complaint. In the case of the neck load we must realise that beside the influence of gravity and high G load, muscle forces are added on to bony structure and soft tissue, so by increasing muscle strength in the cervical area then of course it can shift the problems of muscle fatigue and muscle soreness and it shifts to overloading the soft tissues and bony structures. Do you have any epidemiological for evidence that. What is your comment on that.

Dr Biesemans, Belgium

No we don't have an epidemiologic study. We were astonished to learn that so few of our pilots were doing a physical fitness program.

Dr Landolt, Canada

While Col Schall and Major Vanderbeek are there I would like to ask a question about the age factor. Vanderbeek found it to be dependant but Col Bieseman found that there was no factor. Do you have any comments.

Dr Bieseman, Belgium

Yes but we have only a very small group of course. But in the sample we surveyed we could find no significance.

Major Vanderbeek, USA

Well perhaps it just is that I had a much larger sample and it was only noted in the major injury category. It was not demonstrated in the minor injury category or in the overall summation of any injury versus no injury. It's a fairly soft finding. I'm not sure.

RADIOLOGICAL INVESTIGATION OF THE VERTEBRAL COLUMN OF CANDIDATES FOR MILITARY FLYING TRAINING IN THE ROYAL NORWEGIAN AIR FORCE

by

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INTRODUCTION

Neck injury with associated pain in the cervical spine and its supporting structures of ligaments and muscles are frequent complaints among aviators flying high performance fighter (HPF) air craft. Two recent surveys conducted among air crew of the United States Air Force (USAF) and the United States Navy (USN), respectively, report incidences of neck pain incurred during flight of approximately 50 and 75 per cent in these two HPF pilot populations (Knudson et al 1988, Vanderbeek 1988).
Muscular pain, ruptured ligaments, sliding vertebrae and compression fractures have been described resulting from violent maneouvering during HPF air combat exercises (Andersen 1988). The cervical spinal column carries a heavy load during high-G accelerations supporting the head and various pieces of personal flying equipment. This top-heaviness is expected to increase further with night vision goggles and integrated weapon systems control devices added in an attempt to extend operable conditions. Moreover, since the aeromedical emphasis has been on developments supporting cardio-vascular and respiratory functions, the neck and the delicate structures of vessels and nerves running with it are becoming increasingly vulnerable to damage. Literary speaking, the neck has become pinched between the desire to add weight to the head for purposes other than protection, and, the support to cardio-vascular and respiratory organ systems which allows additional intensity and time spent during excursions into the high-G enviroment.
For these reasons, and because the vertebral column is relatively inaccessible to clinical examination, the medical selection procedures for military flying training with the Royal Norwegian Air Force (RNoAF) were extended some years ago to include a series of roentgen films of the vertebral column with emphasis on the cervicalspine. Our main findings and their consequences for admission to military flying training are presented in this paper.

MATERIAL AND METHODS

During the past 4 years radiological examinations have been carried out on 232 applicants, 221 males and 11 females 19-24 years old. The radiological examination consists of 9 films:
The cervical spine frontal and lateral view with left and right obliques added, the cervico-thoracic area in oblique projection, and, the thoracic and the lumbar spine viewed frontally and laterally. If indicated, functional films with flexion and extension of the cervical column were made. Likewise, "scotty-dog" projections would be indicated if spondylolysis with or without olisthesis were suspected.
The cervical films and that of the cervico-thorecic area are made standing up, those of the thoracic and lumbar spine are taken in the supine position.
The medical personnel involved has been limited to 3 specialists in radiological diagnostic procedures, and 4 technicians all of whom received additional training in order to ensure good quality, standardized films.
One major concern introducing the programme outlined has been to reduce radiation as much as possible by limiting the population exposed. For this reason the radiological examination comes at the end of the medical selection procedure.
The applicants have, thus, passed psychological testing, admission interviews and physical examination, before meeting the Air Force Board of Medical Selection for military flying training. Presently, only those who meet the criteria for flying HPF aircraft are admitted. Consequently, fewer than 10% of the applicants go on to radiological examination. Therefore, it is reasonable to expect that the selection procedure - as a biproduct - provides a basis for determining the normal distribution of radiological diagnoses of the spine in the asymptomatic, healthy population of young adults in the country.

RESULTS

Analyses of the films revealed 527 aberrations, 76 anomalies (A), 81 degenerative changes (B) and 370 aberrations of posture (C), 2.27 diagnoses per x-rayed spine, on the average.
The distribution of positive findings among the three major subdivisions of the vertebral column shows 141 conditions referred to the cervical spine, 173 located in the thoracic column with the remaining 213 deviations appearing on the lumbar films. It appears that anomalies (A) are rare in the cervical column and to some extent also in the thoracic spine, but they are rather frequently seen in lumbar vertebrae. Degenerative changes (B) occur in the thoracic part of the vertebral column with a frequency almost twice that observed in any of the other subdivisions of the spine. Slight to moderate postural changes (C) are evenly distributed among the three main parts of the vertebral column.
The roentgen changes described have been summarized in Table I.
Among significant anomalies it appears that ransitional vertebrae are relatively common and equally distributed between the thoracico-lumbal (9.0%) and the lumbo-sacral area (8.6%). Postural aberrants were frequently observed as well, with slight scolioses and straightening of curvatures affecting a much larger number of the candidates than hyperkyfosis and -lordosis. Degenerative changes of the thoracic spine are largely due to juvenile kyfosis or Mb. Scheuermann in these young adults

with trapezoid vertebrae, reduced disk height and a rugged appearance of the horizontal outline of corpora, all of these changes affecting at least three adjacent vertebrae in order to satisfy the criteria for diagnosis. Moreover, micro-herniation of the nucleus pulposus (Schmorl's nodes) is frequent in the thoracic and the lumbar spine.

Reduced disc height affected almost 18% of the candidates. Spondylolisthesis with or without olisthesis was seen in 12 individuals corresponding to 5.2% of the applicants. The total of numbers tabulated deviates somewhat from those given in Fig 1, the reason being differences between radiological interpretation of visual images relative to their significance for the selection procedure. Since the films were meticulously scrutinized by radiologists well aware of the purpose for which the examination was performed, any however slight, deviation from normal status described has been included in Fig 1. But, if uncertainties of a finding was entertained, differences of opinion expressed as to minute details or if vague language such as "possible normal variants" or "within normal limits" were used by the radiologists, the films have been regarded as negative in this population of otherwise healthy, asymptomatic young men and women. Therefore, Table I and Fig 1 may be looked upon as complimentary by giving the minimax limits of significance of the roentgen changes.

Using this line of argument, it turns out that the film series made from 76 of the candidates would be regarded as negative. Among the 156 others the Medical Selection Board accepted 131 and excluded 25 from military flying training. Twenty were rejected due to roentgen diagnoses alone, the remaining 5 were excluded due to multiple causes among which the radiological excamination was one important factor (Fig 2).

DISCUSSION

Disorders of the vertebral column due to occupational wear usually do not become manifest until middle age. However, frequent and extreme loading of the spine over years as is the case in HPF flying, constitute a chronic strain which may accelerate disease processes of the spine even to the extent of causing sudden incapacitation. Unfortuneately, the correlation between symptoms and roentgen findings is not definite in spinal disorders. This problem becomes additionally augmented when a healthy population of asymptomatic individuals is studied in order to reveal clinically significant roentgen changes or to predict future functional excellence. Thus, the radiological evidence has to be interpreted with great caution. Observing these limitations of the data so far collected, certain guidelines may be useful for the purpose of medical selection, keeping in mind that for the time being the prognostic considerations are necessarily based on knowledge projected from the patient community onto an apparently healthy population.

Because of its great flexibility, the cervical spine is predisposed to injury from forceful movements. Consequently, pathological changes in the cervical column are looked upon as being particularly unfavourable. Therefore, a rather restrictive policy has been pursued in order to avoid candidates with increased risk of incurring a cervical syndrome due to flight in HPF aircraft.

Any anomalies or pathological changes from disease or injury which may contribute to reducing the stability of the cervical spine further or cause a narrowing of the intervertebral foramina or the spinal canal are considered disqualifying. Aberrant curvatures, particularly those which might give rise to increased torque forces are noted and will be added to any other existing deviation described.

Transitional vertebrae which account for most of he anomalies described in the material presented here are accepted although the lever arm of the affected subdivision of the spine become increased. Additionally, the unilateral contact of asymmetrical lumbar sacralization which increases torque forces with consequent strain on the spine and risk of disc herniation above the anomaly is nevertheless acceptable as solitary phenomenon.

No candidate in this highly selected population had developed aberrations of posture to the extent of being rejected for this reason, but, deviating curvatures have contributed to a negative conclusion at final evaluation.

A few candidates, cases of conspicuous roentgen changes, were rejected due to "Scheuermann-changes".

Occupational health programmes have, no doubt, expanded the medical indications for diagnostic procedures which at one time were exclusively reserved for patients suffering from illnes or at least presenting with symptoms suspective of disease. Aviation medicine, for example, is a large consumer of medical services in order to predict and prevent disease among aircrew. Although it is obviously justifiable to perform advanced diagnostic procedures in order to exclude applicants from undertaking a task which might prove harmful, medical intervention may create a problem rather than solve one if the predictive value of the tests precribed is low.

Therefore, a close follow up of populations exposed to examinations is obviously required.

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Roentgen Changes

Anomalies	
Cervical fused vertebrae	2
Thoraco-lumbal transitional vertebrae	21
Lumbo-sacral transitional vertebrae	20
Extra vertebrae	8
Spina bifida	49
Total	101
Aberrations of posture	
Scoliosis	95
Curvatures straightened out	89
Hyperkyfosis /-lordosis	35
Total	219
Degenerative changes	
Spondylolysis /-olisthesis (5+7)	12
Seq Mb Scheuermann	36
Schmorl's nodes	34
Loss of disc height	41
Osteochondrosis	9
Trapezoid vertebrae	16
Previous injury	10
Operated	2
Total	160
Sum Totals	480

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- TABLE I -

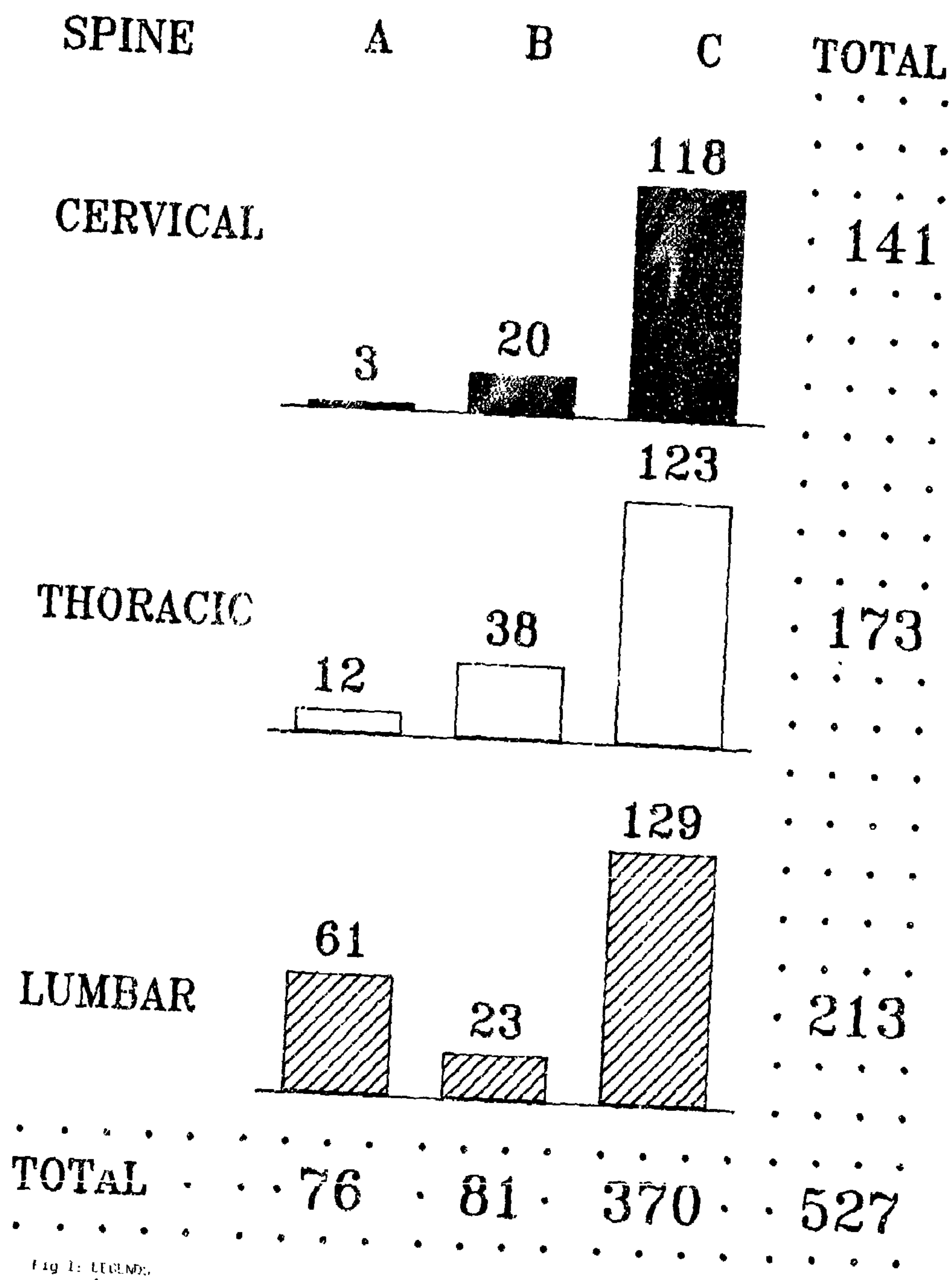

Fig 1: LEGEND:
Anomalies (under A). Degenerative changes (under B) and aberrations of posture (under C) are shown for the cervical, thoracic and lumbar subdivisions of the spine, totals at bottom (horizontal & hatched). Far right column (vertical & hatched) sum total roentgen changes described in each spinal subdivision i.e. cervical, thoracic and lumbar, respectively.

RADIOLOGICAL EXAMINATION CONSEQUENCES

Applicants, total number	232
Negative films – "Normals"	76
Acceptable radiology	131
Excluded by radiology only	20
Excluded, radiology contributing	5

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- Fig. 2 -

DATA ANALYSIS IN CERVICAL TRAUMA

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Summary:
The curvature of the cervical spine in lateral view is discussed and a method based on digital statistical analysis is used to reproduce quantitative data of the curvature. Part I is a study based on the lateral view in the neutral position in 142 aviators. The radiograms are divided in 3 main group: 1) Normal cervical lordosis. 2)Marked straightening of the cervical spine. 3)Segmental straightening with reversal of the curve. Part II discusses the use of the digital analysis to determine the displacement in subjects that have sustained ligamentous injuries of the cervical spine following whiplash injury.

The normal curvature of the cervical subdivision of the vertebral column is a smooth lordosis.
After trauma changes of curvatures are frequently seen, and straightening of the cervical spine is often assumed to be attributed to muscular tension following trauma to the neck.
This seems to be a very common finding, however, a normal curvature usually returns as the symptoms subside. These quantitative measurements of change of curvature correlate well with clinical symptoms.

We have made an attempt at using a method based on digital statistical analysis in order to obtain reproducable quantitative data of the curvature found in the normal group of subjects.

Our re ults have been obtained by applying the digitalizing method to a material consisting of 142 aviators and a smaller group of subjects with documented injuries to the cervical spine.

The digitalized statistical analysis of the base line data has been performed at the Armstrong Aerospace Medical Research Laboratory, Wright Patterson Air Force Base.

The x-rays films that forms the basis for this study have been taken and supplied by the Radiology Department, Oslo Emergency Medical Center, Oslo City Hospital.

The study is reported in two parts:

Part I: A study of changes in the curvature of lateral X-rays of the cervical spine in neutral position is discussed.

Part II: The use of the digitalized method in patients that have sustained ligamentous injuries with instability following whiplash injuries is reported.

Method

The method involves accurately plotting the outline of each vertebrae in the cervical spine and some bony prominencies, giving an outline of the spine. At the same time it has been possible to arrive at accurate measurements between bony structures.
Due to enlargement, in spite of standard distances, subsequent X-rays will be comparable by using a correction coefficient so that fine adjustments can be made by using stable bony structures as land marks to allow for such changes.

The degree or depth of the cervical curvature is a function of the curve produced by the cervical bodies in a lateral view in a neutral position. Such measurements can be obtained by drawing a straight line from the superior posterior aspects of the odontoid process to the posterior inferior corner of the body of the seventh cervical vertebrae.

The line tracing the posterior vertebral bodies will usually produce a crescent shaped line corresponding to the curvature of the vertebrae.

The longest perpendicular between these two lines which usually falls in the vicinity of C4 will give a measurement of the depth of the cervical spine.

In a spine with marked straightening the measurement will be close to zero. In a normal curvature with retained lordosis this value will be positive, in most studies a mean value of 11,8 millimeters. In a spine with a reversal of the curvature a negative number has been reported.

Part I

We have subjected the lateral cervical view in neutral position to digitalized analysis of 142 aviators. All subjects were fighter pilots with logged flying time from various fighter planes including the F-104, F-5 and F-16. We have in this study listed the findings to fit into the three main groups defined above.

We have no information of previous injury, and all aviators were fit for flying. We have assumed that osteochondrosis and degenerative changes are related to the age of the individual. Moreover, we have assumed that the age of the individual is related to the length of the flying time. As in most air forces some senior officers still maintain flying status in the RNoAF.

We have divided radiograms into three main groups:

1) Normal cervical lordosis
2) Marked straightening of the cervical spine
3) Segmental straightening with reversal of the curve of the cervical spine.

Out of the 142 aviators 63,3% had normal cervical lordosis. 26,8% showed straightening of the cervical spine, and 9,9% showed segmental straightening with reversal of the curve. The average age of for these three groups showed no significant difference - 26,4 - 26,8 - 26,1 years as average.

Discussion

This is in accordance with other studies that supports the view that the documented changes are within normal limits and that these changes can not be used as a criterion suggesting pathological changes in the cervical spine.

Part II

Flexion injuries are not uncommon in aviation medicine and has been recorded in accelleration of high intensity from positive to negative G. One of the authors of this presentation sustained such an injury in the passenger seat of a F-16 during a sudden and unexpected evasive manoeuvre.

In this second study we have examined individuals that have been subjected to whiplash injuries. It is assumed that the mechanism of the extension/flexion movement of the cervical spine in this injury is well known.
However, in modern cars the headrests are fixed with an upper border above the level of the external ear. This may to some extent reduce the extension movement and perhaps increase the momentum of the forward flexion of the neck at the time of injury.

It is expected that this trauma may produce injury to the posterior ligament complex including the supraspinous and intraspinous ligament. Rupture of the posterior part of the intraspinous disc with haemorrohage may occur and predispose to later degenerative changes.

When such injuries are suspected a functional study of the cervical spine should be made.

a) The most consistent finding is that the space between the spinous processes at the level of injury is greater than the interspinous distances above and belov the affected level.

b) A horisontal displacement of the adjacent vertebrae in flexion.

c) Marked limitation of movement of segments of the cervical spine is a finding suggesting injury.

Therefore, suggestion of segmental straightening of the cervical spine with reversal of the curvature may be indicative of such a lesion following injury, and a functional examination of the lateral view of the cervical spine should be undertaken.

There is, however, an interesting detail that involves the point of inflection between the two curves formed.

The point of infiection may conincide with the displacement of the apophyseal joints and may be explained in the following manner. Capsular ligaments of the apophyseal joints are dense fibrous structures, providing stabilization and limit the horizontal displacement of the adjacent vertebrae. Therefore, damage to the capsules allows forward displacement of the crainal facet surfaces, with loss of parallelism of the articular surfaces and widening of the joint spaces posteriorly.

These changes are demonstrated using the same digitilized analysis method, demonstrating a displacement as indicated above.

Discussion

The advantages of this method are accurate and objective mesasurements of the displacement of the cervical spine.
Detailed objective measurements between subsequent X-rays studies may be obtained by adjusting for slight degrees of enlargement by compensating through fixed bony land marks.

What is perhaps even more important is the fact that these accurate measurements may prove extremely valuable in producing objective measurements of minute but gradual changes that occur in degenerativ disorders of the spine. This will allow us to study consecutive series of films over a longer timespan and may provide us with an accurate measurement of the development of ostechondrosis in the cervical spine and how this may be affected by trauma or of change that occur due to the forces that the cervical spine is subjected to, for instance by high G forces.

Changes in the cervical spine may be recorded in relationship to time so that we may be able to evaluate the rate of development of ostechondrosis eit

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PROGRESSIVE CERVICAL OSTEOARTHRITIS IN HIGH PERFORMANCE AIRCRAFT PILOTS

by

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SUMMARY

Thirty-one pilots who have been subjected to a repetitive +Gz environment were evaluated clinically and roentgenographically against age and sex-matched controls. Analyses demonstrated significant deterioration in the young pilot groups compared to controls in terms of neck range of motion, osteophytic spurring at C5/C6 and disc space narrowing at C4/5 and C5/6. While the pilots remain relatively asymptomatic during their flying career, they may be at greater risk for symptomatic cervical disease later in life. The +Gz environment appears to play a role in an accelerated rate of cervical osteoarthritis in high performance pilots. This must be taken into consideration when systems that will increase the longitudinal impact load on the neck are being contemplated.

There is an increasing need to evaluate the progressive nature of musculoskeletal disorders affecting the neck. In our practice, we have seen several retired pilots who have presented with symptomatic cervical osteoarthritis with no history of specific neck trauma on or off duty. This led us to wonder whether repetitive +Gz forces applied to the neck during many years of flying high performance aircraft might result in a progressive cervical osteoarthritis.

This paper attempts to establish that progressive cervical osteoarthritic changes occur in pilots exposed to a repetitive +Gz environment as compared to an age matched control population with or without clinical manifestations.

There is a reluctance in the Canadian Forces pilot community to see the flight helmets increased in weight. They appear to feel that such a weight gain will decrease the comfort of the helmets and increase the problems with neck pain.

MATERIALS AND METHODS

Thirty one pilots were evaluated with an age range of 32 years (23 years to 55 years) and a high performance flying experience ranging from 240 to 7,200 hours. Pilots from various age groups volunteered for the study after being given assurance that the medical data obtained would be used for statistical purposes only and would not allow us to identify an individual participant. Individual radiographic and clinical findings were discussed with pilots at the conclusion of their participation in the study.

Fifteen controls who were age and sex matched were also evaluated. The controls were also members of the military, but did not fly aircraft. For purposes of radiographic analysis the data was also compared to a larger age and sex matched control available from the literature (1).

Clinical history and physical examination was carried out by one or the other of the authors using a uniform checklist technique requesting wherever possible a yes/no response or normal/abnormal physical finding. Neck range of motion was graded as 0 for normal, -1 for 25% decrease, -2 for 50% decrease, -3 for 75% decrease and -4 for total lack of motion measured in each of six planes; flexion, extension, right and left rotation and right and left lateral flexion. Both history and physical components of the clinical exam were scored out of a possible 50 points. Abnormal findings resulted in a lower score that is, a normal history and physical exam would result in a perfect score. Abnormalities were scored by deducting points from the total score. History of neck incidents were recorded as total number of incidents during or within 48 hours of a flight. Non-flight related neck incidents were also recorded. Individual observations of the pilots with respect to neck incidents and history were also recorded and will be summarized in narrative form.

ROENTGENOGRAPHIC ASSESSMENT

AP and lateral cervical views were taken of all participants using a consistent radiographic technique. The body of C7 was included in all films. X-ray films were read separately by two radiologists who were blinded as to whether the X-ray films were of pilots or controls. The X-ray films were then evaluated for the following parameters; cervical curvature, paravertebral soft tissues, vertebral body height at C3 to C7, osteophytic spurring at C3 to C7 and disc space narrowing at C3/4, to C6/7. Each criterion was assigned a designation of normal, mild, moderate or severe with abnormalities being awarded a graded negative score resulting in lower overall scores. Each category was tabulated numerically. Total possible correct score was 50. Inter-relator reliability was evaluated between the radiologists and was found to vary betweeen 0 and 8%.

Chi square and "t Test" analysis did not demonstrate any significant difference between the radiologic evaluators.

Statistical evaluation of pooled history and physical results as well as selected age subgroups were carried out using the Student's "t test". Variance analysis was also carried out on pooled and selected subgroups of radiographic findings to test for significant differences in soft tissue or degenerative changes between control and pilot age subgroups as well as between pilot age subgroups. The age ranges used for subgroup analysis were 20-29 years, 30-39 years and 40-55 years.

RESULTS

The total number of high performance jet flying hours was 83,235 with a mean flying experience of 2685 hours ± 1923 hours (± 1SD). As might be expected there was an increase in the number of flying hours with age. (Table 1). The mean ages of the pilots vs controls show a good match in the 30-55 year subgroups. Unfortunately not enough controls were acquired in the 20-29 year subgroup. (Table 2). For purposes of effective evaluation our pilot date was compared with age and sex matched data previously published (1). Comparing our control data with the age and sex matched published control data showed no significant difference. This suggested that the pilot data could be effectively compared against the published control data.

Table 1: Pilot Flying Demographics

Age Group	Number	Mean Flying Hours	± 1S.D.
20-29	8	630.6	± 489
30-39	12	2329.2	± 691
40-55	11	4567.3	±1725
All	31	2685.0	± 1923

Table 2: Pilot/Control Demographics

Age Subgroup	Number	Mean Age (Years)	± 1S.D. (Years)
20-29 Pilot	8	25.1	2.5
Control	1	24.0	-
30-39 Pilot	12	34.6	2.5
Control	7	33.7	3.6
40-55 Pilot	11	47.5	7.0
Control	5	45.0	3.0
All ages Pilot	31	36.7	10.0
Control	13	37.3	7.5

The history data did not show any significant difference between the pilot and control groups either as pooled data or in various subgroups. There was a suggestion that complaints of neck and periscapular pain as well as symptoms of neck stiffness and crepitus may have been reported more frequently among pilots versus controls but this trend was not statistically significant.

There were virtually no volunteered complaints of sensory, motor or vascular changes suggestive of a radiculopathy. No history of posterior circulatory symptoms or gait or bladder symptoms suggestive of a myelopathy were found in this population.

Five pilots (16%) in the sample have had an aircraft mishap involving either a forced landing or ejection. All had some incident of neck discomfort but it was self-limited in all cases. A total of 22 (71%) of the pilots related an experience of neck symptoms either once or many times lasting from hours to months. However in all cases except two, the symptoms were self-limiting. The most common situations leading to neck pain appear to be the following:

1 Air Combat Missions (ACM). In this situation especially in defensive manoeuvres, repetitive pulling of +G in a 5.0 to 8.5 G envelope is carried out over a period of time. This is usually associated with either maintaining or moving the neck in extension/lateral flexion during the +G turn in order maintain visual contact. Most pilots are well aware of a phenomenon referred to as "ACM neck". This refers to neck pain experienced during or after an ACM flight. An exercise program of neck range of motion and isometric neck strengthening is carried out to minimize these symptoms. Again in virtually all cases the neck symptoms are self-limiting.

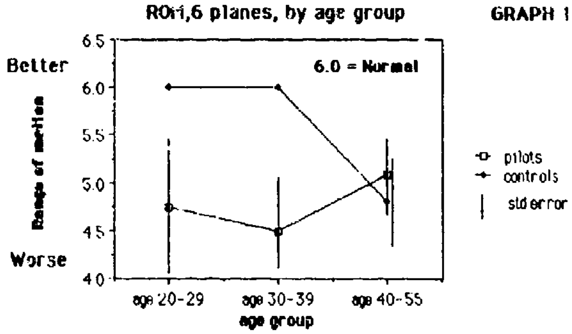
2. "Passenger" Situations. This seems to cause the most serious subjective sequelae and was commented on by pilots and navigator. When an aircrew is riding as a passenger there are situations where +G is pulled by the "driver"

and not anticipated by the passenger. This has resulted in the passenger sustaining a sudden longitudinal impact to the neck without benefit of muscular tension providing protection. It takes the neuromuscular apparatus at least 75 milliseconds to prepare reflexively for an impulse load and absorb energy by muscle lengthening.(2). The pilot knowing that he is going to initiate the manoeuvre has time for the protective response to take place.

3. Aircraft Mishaps. Not unpredictably, ejections and forced landings formed the next most common cause of neck symptoms. Of the pilot sample there were two cases of cervical radiculopathy as demonstrated by EMG changes. Both of these cases were in air crew over the age of 50 years.

The physical examination component for the most part also did not demonstrate significant differences between the pilot and control groups. Sensory examination to pinprick, light touch and vibration (256 Hz) as well as detailed upper extremity motor exam looking at muscle bulk and static strength testing failed to demonstrate any significant difference. A positive Horner's test was not seen in any study participants and Spurling's test was negative in all cases except for the two cases of documented cervical radiculopathy. Upper extremity reflex assymmetry was seen in a few pilots and controls but did not demonstrate significant difference as a possible differentiating tool between pilot and control samples.

The only examination process demonstrating significant difference ($p \leq .05$) was in neck range of motion. Pilots in the 30-39 age group had significantly reduced right and left lateral neck flexion compared to control (two tail t test). This significance was carried through evaluation of all six planes of range of motion but was made up primarily of the lateral flexion components. (Graph 1). The maximum neck range of motion loss seen in our sample group was 75%.

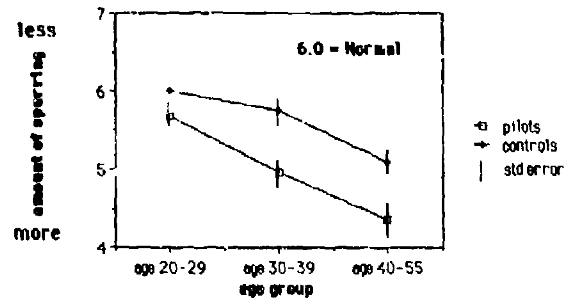


ROENTGENOGRAPHIC FINDINGS

Five parameters were evaluated on the AP and lateral cervical films. In this study, no significant difference was seen between controls and pilots for vertebral body heights or paravertebral soft tissue. Both of these parameters were essentially normal in all participants. In terms of loss of the normal cervical lordosis there was evidence of this occuring in the pilots. However due to the lack of controls in the young age group we were unable to carry out a correlative analysis between pilots and controls. An interesting trend, although not analyzable, was that loss of the normal cervical lordosis seemed to occur more frequently in the younger actively flying pilot groups and that the cervical lordosis was closer to normal in the older pilot group made up predominantly of pilots not currently flying.

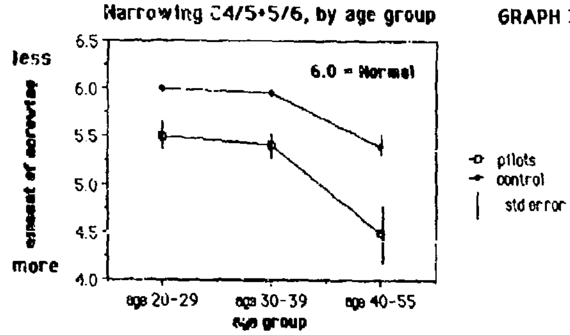
The parameter of osteophytic spurring did show a number of interesting findings. There was no significant difference between pilots and control groups at C3 or C4 at C7. However at C5 and C6 there were marked differences. The pilot group at 30-39 showed significantly more osteophytic spurring at both levels than did either of the control groups ($p \leq .05$, t test 2 tail). Of interest, although the pilot 40-55 age group showed more changes than the 30-39 age group the degree of difference between the pilot and control groups at the 40-55 age group was not nearly as marked ($p \leq .10$). While there was an overall radiographic deterioration with age, the rate of deterioration was more marked for the pilot during the fourth decade and more marked for the controls during the fifth and sixth decades. (Graph 2)

spurring at levels C5+6 by age group **GRAPH 2**



Disc space narrowing showed similar trends. There were no significant changes noted at the C3/4 level in any of the subgroups. At the C6/7 level, some disc space narrowing was noted with increased age but there was no significant difference between pilots and controls at any of the age subgroups. At the C4/5 and C5/6 level there was a strong significant differance between pilots and controls all age groups ($p<.01$) demonstrating more disc space narrowing among the pilots versus control groups. The study demonstrated significant differences especially in the younger age groups 20-29 and 30-39 ($p<.05$). The control groups had virtually no disc space narrowing present at these levels as opposed to the pilot groups. The differences betweeen the pilot and control group was still significant in the 40-55 age group but the trend was not quite as marked. Both 40-55 age groups demonstrated evidence of increased disc space narrowing with age at these levels.(Graph 3) The findings of disc space narrowing most prominant at C4/5 and C5/6 as well as osteophyte changes at C5 and C6 are consistent with the findings of other studies (1,2,3).

Narrowing C4/5+5/6, by age group **GRAPH 3**



DISCUSSION

Pilots of high performance aircraft are subjected to a number of unique environmental stresses as a result of a repetitive +Gz environment. One of the particular long term hazards of this environment would appear to be an accelerated cervical osteoarthritis starting at a surprisingly young age. The fact that the rate of changes are most pronounced in the young age groups who are currently flying and less pronounced in the older non-flying population group also suggest that the changes seen are due to the flying environment. It may also suggest that the rate of changes slow down once the pilot is no longer flying. The other finding of this data is that the changes are not resulting in significant clinical manifestations during the active flying portion of the pilot's career with the exception of diminished cervical range of motion. Thus the pilots are unaware of the cervical deterioration taking place until somewhat later in life, if at all.

Articular cartilage is highly vulnerable to impact loading. This accounts for the high frequency of osteoarthritis related to joint overload as in shoulder and elbows of pneumatic drill operators and baseball pitchers, ankles of ballet dancers, metacarpalphalangeal joints of boxers and knees of basketball players. (2). This would also account for the changes seen in the necks of high performance pilots.

The major forces on articular cartilage results not from weight bearing but from contraction of the muscles that stabilize or move the joint. The major factors that attenuate shock delivered to the joint appear to be joint motion and the associated lengthening of muscles under tension as well as deformation of the subchondral bone under load (2).

The two major etiologic factors resulting in cervical joint degeneration in aircrew of high performance aircraft would appear to be 'unanticipated' impulsive loads and muscle fatigue. As discussed earlier aircrew in a "passenger" role may be subjected to a sudden impact loading without the benefit of protective neuromuscular preparation for shock absorption. The deleterious effect of this has been extensively studied in passengers sitting in motor vehicles involved in low impact rear-end collisions. (2,4,5) Situations leading to cervical muscle fatigue such as repetitive +Gz forces over a period of time will also impair the shock absorbing mechanism. Active motion under control cannot cause a sprain injury, but passive motion with force, either sudden or sustained can produce injurious effects on the joint structures.(4). Excessive loads cause microfractures of the subchondral trabeculae. These changes in subchondral bone produced by accumluated microtrauma affect the ability to absorb the energy of longitudinal impulse loading of the joint and may thereby lead to cartilaue degeneration(2). White (5) in his extensive review of "whiplash" injuries indicates that acute soft tissue cervical injuries with later osteoarthritic changes are typically produced in motor vehicle rear-end collisions where 5 to 15 G of horizontal acceleration is applied.

Jackson (6) points out that C5/6 and C4/5 are more vulnerable to stress/strain and injury. Osteophytic spurring is found most frequently at C4,C5 and C6 (1,6,9). This is certainly consistent with the findings in our study.

It is well known that cervical osteoarthritic changes occur with increasing frequency with increasing age in symptomatic and asymptomatic populations. (1,2,7,8,9). One reference (9) places the incidence as high as 80% by age 55. As well lateral neck flexion especially in males decreases with age (10). Radiographs do not indicate the extent of the actual pathology.

It does not seem that these changes are the result of intrinsic senescence of articular cartilage but rather an accumulation with time of an increasing number of mechanical insults to the joint (2,11). Articular cartilage exhibits a fairly consistent biochemical composition after skeletal maturity with no lessening of the metabolic activity of the cells. However the ability to withstand fatigue testing diminishes progressively with age. (2)

Lieberman (9) points out that bony changes are only part of the picture in cervical spondylosis and radiculopathy. One also requires biomechanical effects of altered cervical motion as well as circulatory factors. As well one requires a narrow spinal canal for symptomatic cervical spondylosis.

While we cannot prove that aircrew in this environment are more likely to develop symptomatic cervical conditions, this study does demonstrate that the observable degenerative changes are significantly accelerated in a young pilot population. It seems reasonable to assume that this creates an environment where symptomatic cervical conditions are more likely to occur.

As the force of impact loading is a function of the number of Gz and the weight of the head/helmet system, then increasing either of these parameters will increase the repetitive longitudinal impact load on the neck. This will likely result in an increased rate of degenerative cervical changes and perhaps result in an increased incidence of symptomatic necks either acutely or long term.

In light of this caution is advised when considering increasing the weight of the aircrew helmets for high performance aircraft.

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DISCUSSION PERIOD 2

Dr Landolt, Canada.

Dr Andersen. It's almost three years ago that you had that neck injury in the F16B, are there any lingering or residual signs of that injury.?

Dr Andersen, Norway.

Yes. I have a very tender spot where the ligament ruptured. There is no doubt that particular injury is staying with me.

Dr Bishop, Canada.

I would like to ask Dr Gillan a question. I was rather impressed with your notion of the changes in the neck of pilots flying high performance aircraft, as an example perhaps of what is known in industry as repetitive strain injury. What suggestions might you have for reducing the incidence of these in light of some of the things done in industry to reduce repetitive strain injury.

Dr Gillan, Canada.

That is an extremely good question. I think that there are a number of ways from the physicians point of view that you can look at that. One is to try and minimise or avoid the risk parameters that go into the problem in the first place; such as the weight of the helmet, as well as the G envelope that is required. That particularly is unlikely to be changed. A number of the pilots, and that was alluded to, in this mornings paper had began to identify corrective techniques to bring their neck into a more optimal parameter. The one that was described to me was, especially in the air combat manoeuvres, to try not to look at the bogey all the time, but rather to take a fix, reposition your head, pull G, relocate. That is certainly healthier from the point of view of the neck but I can't comment on the survivability in that situation. It is certainly better for the neck if it is possible to maintain visibility, at the same time as trying to translate some of the force parameters, for instance from the occipital area down to the shoulders, and bypass the neck. However I can't comment on whether that engineering feat is actually possible or not, it would certainly be interesting to investigate, and one of the papers this morning, looks at some of these possibilities

Col Hickman, USA.

Dr Gillen, your paper will undoubtedly be widely quoted because of the implications for acceleration of degenerative disease in fighter pilots and in that regard it has enormous impact in occupational medicine. With a sample of 31 and with over half a dozen variables I am concerned about the statistical sample required and the power that is available for you to make these conclusions. In particular I would like to ask whether or not these are radiographic criteria for which you have stated percentages or measurements. I would urge you to make sure that everyone understands what percentage you would have considered constitutes a significant difference. For instance in vertebral height, what constitutes a significant difference, because that would constitute the sample size and I believe it is incumbent on us, before these studies are widely quoted, to make sure that the sample size will support an argument of this size especially where percentages are involved. You could easily state that you think 20% is significant or we think 1% is significant, because then it would dictate the sample size. It is a well put together paper. I am concerned about the sample size and the conclusion based upon this power.

Dr Gillen, Canada

I am seeking a much larger sample. If I can embellish upon that, I certainly agree with everything you say. There are a variety of stages that can be done with clinical studies. The first step is to do a sampling. If the sampling is large enough to give you some statistical validity, well and good, but a sample of this size is merely that, its a sample. It was interesting but even using low number variance analysis we were able to pull out statistical significance, but I would agree that the interpretation should be merely to provide the impetus for a much larger study. The study was trying to do a number of things: 1) To demonstrate whether there was something there. 2) It was to attempt to standardise some techniques of quantifying looking at a clinical situation. On the basis of this we were attempting to standardise and quantify radiographic techniques as well as a history and physical.

ELECTRONYSTAGMOGRAPHIC FINDINGS FOLLOWING CERVICAL INJURIES

by

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Introduction

The cervical column consists of seven vertebrae with thirty-eight joints. The head weighs about 4600 grams; however, the musculature of the neck adds considerably to the pressure on the vertebral column by tension of the neck muscles for they function as the stays of the mast of a ship. This adds up to 50% of the total weight.
Head movements are made by changing the alignment of the cervical vertebrae that link the skull with the body. The movements are controlled by the activities of more than 20 pairs of muscles that link the skull, the spinal column and the shoulder girdle. The biomechanical model of the head-neck system pursuits a whole spectrum of movements in all planes of motion and for a variety of speeds. The muscle actions are constrained by the physical properties of the vertebral column, whose articulations differ in their ranges and directions of mobility. This model has three types of elements:

- the flexible beam: the vertical column down to the upper thoracic area, including the soft connective tissues,
- the rigid mass of the head,
- the neck muscle complex, divided in:
  - the posterior muscles (m. semispinalis capitis),
  - the anterior-transverse muscles (m. sternocleidomastoideus complex),
  - the posterior transverse muscles (m. splenius),
  - the lateral muscles (m. longissimus capitis).

Origins and insertions, axial rotation and muscle tensions can be approximated easily, so that the resulting peak static torques are compatible with the known torques generated voluntarily in various directions.
All types of rotational movements can be performed with this construction, while the axis of rotation changes automatically with changing conditions. Also transitional or sliding movements will occur automatically. Most head movements are made while using several muscles. The coordination of these muscles is not clear, as several neck muscles can share a similar pulling direction. The particular role played by a single muscle cannot be predicted by its pulling direction.
The neck muscle system has eight neural controller locations and eight locations for sensory feed-back.

With regard to the vestibulo-spinal connections two major projections are known:

the <u>lateral vestibulo-spinal tract</u>, originating from the lateral vestibular nucleus (Deiters), arises from neurons which receive input from the anterior vertical semicircular canals and from the otoliths, and project to and excite homolateral dorsal neck motor neurons.

the <u>medial vestibulo-spinal tract</u> contains multiple components, originating from all three canals and otolith organs. The horizontal canal input excites contralateral neck motor neurons and inhibits homolateral neck motor neurons. The anterior canal input excites the contralateral dorsal and ventral neck motor neurons and inhibits homolateral ventral neck motor neurons. The posterior canal input excites ventral neck motor neurons and inhibits homolateral dorsal and ventral neck motor neurons and contralateral dorsal neck motor neurons. Some neurons also mediate otolith input to inhibit homolateral neck motor neurons and to excite as well as inhibit contralateral neck motor neurons.

Neurons, both from the lateral vestibulo-spinal tract and the medial vestibulo-spinal tract which project to the neck motor neurons also send branches to other levels of the spinal cord and to brainstem nuclei including the oculomotor nuclei.
The motor system of the neck can be considered as abundant as it can produce the same movement using an infinite number of different patterns of muscle activation. This means that the system uses some types of optimality criterion in choosing the particular pattern

Passive rotation of the head ensues, after a short delay, a rapid eyemovement in the direction of the head rotation. This movement originates from the semicircular canal system
To attain the visual target the gaze is controlled by coordinated eye-head movements. In order to control the position of the visual axis precisely, the eye and head positions are monitored by corollary discharges that calculate an internal representation model of the current gaze position. This is compared to the desired gaze position to yield an internal signal specifying gaze position errors. Tecto-reticulo spinal neurons are responsible for these gaze signals. The superior colliculus plays an important role in the coordination of extra ocular and neck muscle commands.

It is clear that the complicated network of muscles and neural mechanisms makes the cervical area vulnerable to traumata.
Among the blunt traumas to the cervical region are the acceleration-deceleration accidents, better known as cervical whiplash injuries.

Since the first report of Gay and Abbott in 1953 about the effect of acceleration traumata on the cervical vertebral column, this type of injuries has drawn considerable attention.
The invention of traffic lights, an increasing use of automobiles and traffic jams enhanced the incidence of accidents, especially the rear end collisions.
When the human body in sitting position is exposed to a blow from the rear the trunk accelerates for a short while, whereas the head lags behind. When the head strucks the head support, which limits the extension of the head, the next moment it is thrown forward which results in a rather strong flexion. When the head reaches maximal flexion a swinging force stretches the cervical column. The muscles need 5-20 m sec for activation. So the movement is completed before the muscles are able to give any resistance, as the incident lasts less than 50 m sec. That is the picture of the mechanism of the cervical whiplash injury.

<u>Material</u>

In the vestibular department 173 patients, suffering from the consequences of an acceleration accident of the neck have been investigated during the last 2 years. These patients were aged from 4 till 58 years. All patients acquired the trauma at an automobile accident. The patients visited the department because of persistent complaints as headache, dizziness, tinnitus and visual disturbances. In all patients an ENT-investigation, audiometry, vestibulometry and visual tracking tests were performed.
Although all patients had the type of injury in common, combinations of complaints differed considerably.
The complaints are given in table 1.

<u>173 patients with cervical whiplash injuries</u>

unconsciousness	52	30 %
headache	152	88 %
cervico-brachialgia	163	94 %
vertigo - dizziness	136	79 %
tiredness	118	68 %
memory difficulties	54	31 %
difficulty in concentrating	49	28 %
depression	38	22 %
irritability	16	9 %
tinnitus	62	36 %
visual disturbances	42	24 %
hearing disturbances	21	12 %
decreased alcohol intolerance	28	16 %

table 1

When unconsciousness for more than 15 minutes following the accident for more than 15 minutes occurred the development of psychological problems appeared to be two times higher.
The most common complaint was cervico-brachialgia which appeared in 94% of the patients. Headache with 88% and vertigo with 79% are well represented too.
All these complaints did not show a consistent character. Free intervals and changes in severity were rather common with regard to all types of complaints.
Headache in most cases was radiating from the frontal region (17%), from the occipital side (59%) or from the vertex (24%). This headache was chronic in 24% of the cases, paroxysmal in 32%, and its presence was shown in 44% when provoked by activities.
Vertigo was present in 79% of the patients. In 47% of these patients the vertigo was of a chronic type, in 35% of a paroxysmal type, which means that vertigo appears for periods of 5 minutes to several hours, while in 18% vertigo only appeared during head movements. In the first two types the vertigo changed severely in most cases over the time, sometimes with free intervals of several weeks.

<u>Examination</u>

Examination of the patients revealed that only 28% suffer from some limitations in head-neck movements.
Keeping in mind the sophisticated structure of the cervical musculature and cervical vertebral articulations the discrepancy between the presence of cervico brachialgia and movement restrictions can easily be explained. Although in all patients X-rays of the cervical vertebrae were made, only in a few cases pathology was found which could be linked to the accident.
A spontaneous and positional nystagmus of more than 5°/sec was found in 60% and 61% of the cases respectively. In a normal population this sign of vestibular system pathology appears in less than 2% of the cases. (See table 2).

173 patients with cervical whiplash injuries

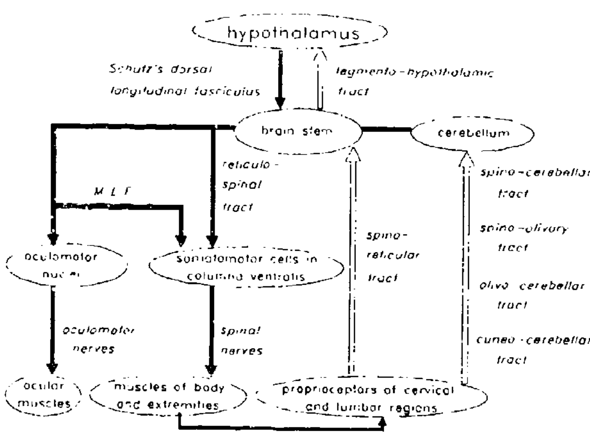
limitation of neck rotation	48	28 %
cervical induced nystagmus	134	77 %
spontaneous nystagmus	104	60 %
positional nystagmus	106	61 %
bilateral gaze nystagmus	138	80 %
visual pursuit disturbances	142	82 %
optokinetic nystagmus pathology	70	40 %
peripheral vestibular lesion	8	5 %
no E.N.G. pathology	8	5 %

table 2

A cervical nystagmus examination was done by successive head rotation to the right and to the left for 30 sec. A cervical nystagmus can be of a proprioceptive type or a vascular type. In 77% of the patients a cervical nystagmus was found, of which in 80% of a proprioceptive type which means lesions of the cervical neural roots.
A bilateral gaze nystagmus is a sign of brain stem pathology. This nystagmus is found when a gaze deviation of 30° is maintained to the right and to the left side successively for a duration of 20 sec. In 80% of the cases this pathology was found. Also pathology in the visual pursuit movements was found in the same patients which points to brainstem and cerebellar pathology. The visual suppression test during the rotation test, evident for cerebellar pathology, was positive in 30% of the cases.
This means that in 82% of the patients lesions were apparent in cerebellar and brain stem structures.

Discussion

It is characteric that the described patients were more disabled and remained much longer handicapped than was anticipated, considering the mild character of the accident. The described cases came to our department in a period of 6 months following the accident, which means that only patients with persistent complaints were investigated.
The findings suggest that overexcitation of the cervical proprioceptors may be an etiological factor in causing vertigo due to whiplash injury.
The abnormalities in the visual pursuit movements indicate a dysfunction of the central nervous system.
During a cervical whiplash injury the flexion of the head means a pull on the cervical medulla, which pull is conducted up to the medulla oblongata and to the brain stem. The pull can mean a lengthening of these structures of up to 5 cm.[2]
It is clear that such a pull can cause extensive damage to the central nervous system[1].
Long after a whiplash injury the lesions to the cervical soft tissues and the ruptures of muscles and ligaments have changed morphologically to inflammatory granulation tissue with scarring and degeneration of nerves in the cervical area[4].
Our findings that 79% of the patients suffer from vertigo is rather similar to the report of Hinoki[3] who found vertigo in 87% of this cases. According to this author vertigo develops according to table 3.



Neural Mechanism related to Vertigo due to Whiplash Injury

table 3

Cervical whiplash injuries cause damage not only to the cervical soft tissues and ligaments, but also to the central nervous system, often in such a way that persistent damage is done. Head restraints support the head against the threat of overextension, but flexion most probably gives far more damage. This means that the protective effect of head restraints is limited, which was reported already by States et al. in 1972[5]. The reduction of the incidence of neck injuries was disappointing too. When the head, as the most vulnerable part of the body, has to be protected against acceleration accidents head restraints have to surround the head. Perhaps a challenge to designers of automobiles.

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AIRCREW NECK INJURIES A NEW, OR AN EXISTING, MISUNDERSTOOD PHENOMENON?

by

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The first U.S. Navy ejection occurred on 9 August 1949. Since then, the Navy has experienced 4,335 ejections through March 1988.

The first serious neck injury, a paracervical strain, associated with an ejection occurred on 20 February 1953. That injury, also, was the Navy's first vertebral/paravertebral injury associated with ejection. Initially, assessment of such neck injuries was accomplished as simply another vertebral injury, however, in more recent years we have become increasingly aware that the cervical/paracervical injuries reported in connection with ejections do not exhibit the same pattern characteristics as other vertebral/paravertebral injuries. They occur at different rates than would be expected based upon the rates of injury for other vertebral regions. Further, although a clear relationship between through-the-canopy ejection and higher incidence rates of vertebral injuries had been established (and a causal mechanism identified), that appeared not to be the case with cervical/paracervical injuries. In addition, while vertebral/paravertebral injury incidence rates have fallen with the changes in catapult boost acceleration, cervical/paracervical injury incidence rates have steadily increased.

Why are cervical/paracervical injuries associated with ejection increasing; why have they not declined as the injury incidence rates have for other spinal regions? Anecdotal and statistical mishap data examined and compared in this and prior studies by the authors suggest that the answer might lie in the aircraft maneuvering capabilities and in an increased frequency of ejection during, or following, gyrations resulting from loss of control of aircraft. The answers we have obtained suggest that considerable caution needs to be exercised in the current trend to integrate manifold systems elements into aircrew helmets, especially in light of the "man-limited" maneuvering capabilities of today's modern tactical aircraft.

INTRODUCTION

On 26 September 1974, the Naval Safety Center, Norfolk, expressed concern in a naval message that the frequency and severity of neck injuries sustained by ejectees were increasing significantly. This message touched off a flurry of activity, much of it still continuing, aimed at assessing the claim and ascertaining the causal mechanism(s) and factor(s) for "ejection associated" neck injuries. The neck injuries of particular interest then and now are:

- Cervical Transections
- Cervical Subluxations
- Cervical Fractures
- Paracervical Sprains
- Paracervical Strains

This paper is one of a series of midstream reports concerning our progress and findings in probing the U.S. Navy ejection statistics. In our earlier papers we have limited ourselves to the assessment of data presented in the Naval Safety Center, Norfolk, ejection mishap data tapes. These data, covering the period from 1 January 1969 through early 1988, were encoded and synopsized by the Center from MORs/FSRs (Medical Officer's Reports/Flight Surgeon's Reports) prepared and submitted for each ejection mishap and presenting data concerning the aircrew and the aircrew life support systems (ALSS) equipment, i.e., helmets, flight suits, personnel parachutes, ejection seats, etc.

These data, however, proved inadequate for ascertaining whether there has in fact been any significant trend affecting the frequency and/or the severity of "ejection associated" neck injuries.

Accordingly, in this portion of this on-going effort, in an attempt to ascertain whether the current frequency and severity of "ejection associated" neck injuries differ significantly from those of U.S. Navy earlier escape systems, we have explored the limited ejection data contained in a log of 1,968 U.S. Navy pre-1969 ejections and ejection attempts (beginning with the first U.S. Navy ejection occurring in 1949 and ending 31 December 1968). In addition, we have compared, and are comparing, the data contained within the pre-1969 log to the data contained in the Naval Safety Center, Norfolk, ejection mishap tapes (**Figure 1**).

Although the pre-1969 ejection log data was utilized, it was used cautiously for the data base is beset with a number of problems, especially among the data for the first decade of the log:

- Imprecise Data
  - Difficult to Interpret Ejection Altitude Data in Terms of System Terrain Clearance Performance Capability
  - Contains Many Ambiguous and Non-Specific Terms Which Require Interpretation
- Total Available Data is Limited, Especially for the First Decade.
- Escape System Modifications Not Always Recorded, e.g., ZDL (Zero Delay Lanyard), in Manner Permitting Determination Whether Present and Used.

DATA PREPARATION

The first step in attempting to compare these data was to review both sets of data to eliminate those cases, both those with and those without neck injuries of interest, which were either not accomplished clear of the aircraft or in which the ejection was initiated outside the system's terrain clearance performance envelope. The resulting populations of ejectees (**Figure 2**), therefore, except for those experiencing major system failure resulting in non opening of the personnel parachute, do not include ejectees with multiple extreme impact injuries, a condition which makes virtually impossible the rational determination of an injury's causation, i.e., was it induced by ejection forces, windblast, crash loads, or unretarded body impact with the surface, etc. [A few cases in the Naval Safety Center's data tapes have been classified by the MOR/FSR preparar as "out of envelope" based upon the ejection airspeed. To the extent that we were able to identify those cases, they have been added back to the data base since they might help anchor and define the role(s) of airspeed in producing these injuries.] Neck injuries among ejectees experiencing non-retarded surface impact as a consequence of personnel parachute failures were not examined.

To the original list of neck injuries of major concern, we have added the following in this effort:

- Cervical Compression Injury Without Fracture
- Cervical Crushing
- Cervical Disc Ruptures
- Cervical Disc Hernias
- Stretching of Paracervical Muscles
- Tearing of Paracervical Muscles/Ligaments

This expansion, while adding no new cases to the ejectees populations previously considered, helped provide assurance that the population derived from the tape data base was as complete as feasible. None of the pre-1969 log data included any notations of any of the above listed injuries either.

DATA SORTINGS AND ANALYSES

As shown in **Figure 2** and **Table I** there were 1,764 ejections recorded in the pre-1969 log which appeared to meet our criteria for being accomplished clear of the aircraft and within the escape system's terrain clearance performance envelope. For the period covered by the mishap tapes, 1 January 1969 through early 1988, the number is 1,677

These two data bases were searched (the pre-1969 data base manually and the post-1968 data base by machine) to identity all recorded neck injuries of the types described in the two lists above.

Twelve of the pre-1969 ejectees (0.68%) were reported to have sustained cervical fractures, while 28 of the post-1968 ejectees (1.67%) were reported to have sustained similar injuries. [One pre-1969 A-5A ejectee lost at sea was reported to have descended hanging motionless beneath his parachute, a condition reported for the one pre-1969 A-5A and two post-1968 RA-5C ejectees receiving fatal neck injuries However, since the suspected injury could not be confirmed, the ejectee is not included in any neck injury population.]

Paracervical sprains and strains were reported to have been sustained by 113 (6.41%) of the pre-1969 ejectees and by 204 (12.16%) post-1968 ejectees. It is important to recognize that the terms "sprain" and "strain" are employed in the post-1968 data base to describe a wide range of symptoms, e.g.:

- Paracervical Stiffness
- Paracervical Soreness/Tenderness
- Paracervical Spasming

Accordingly, we currently are engaged in an effort to classify these sprains and strains by severity, an especially difficult task for the pre-1969 data base. This requires development of a common basis between the pre-1969 and post-1968 data bases for judging the injury severity and/or the acquisition of additional pre-1969 data. Both approaches are being pursued.

Table I summarizes the annual pre-1969 in-envelope ejection data, providing the annual and total number of ejections, ejectees receiving cervical fractures, and ejectees receiving paracervical sprains/strains. **Table II** provides similar data for the post-1968 period. **Table III** displays the pre-1969 data by aircraft model and type of ejection seat, while **Table IV** provides a similar display of the post-1968 data. The periods during which each of these aircraft models appear in the pre-1969 log and in the post-1969 data base are shown in **Tables V** and **VI**, respectively.

The pre-1969 data proved incomplete concerning canopy mode of the escape with large numbers of cases having no mode shown. Even after inserting values for those systems in which there is no flexibility, there remains, with the currently available data, a large body of unassignable cases, precluding assessment of the role(s) of canopy mode in the production of these neck injuries among the pre-1969 ejectees.

Likewise, there is a paucity of pre-1969 data with which to evaluate the role of aircrew preparedness for the onset of G forces in the production of these neck injuries. The only available pertinent data is the presence or absence of powered haul-back type inertia reels in the seats. These first appeared with the introduction of the LS-1 and HS-1 ejection seats and, later, gradually appeared in ESCAPAC ejection seats and then MK7 series ejection seats.

In the pre-1969 logs there is some data concerning collisions with trees, buildings, etc., during parachute descent, concerning parachute landing falls, and concerning the presence of prior neck injuries. There is not, however, any anthropometric data for the ejectees. There also is a limited amount of data concerning ejection seat malfunctions which might have influenced the neck injury incidence rate.

SUMMARY ASSESSMENTS

From this effort and our earlier analyses, we have developed an initial feel for the data which suggests that there is no single neck injury causal factor or mechanism but, rather, that the causes of "ejection associated" neck injuries are manifold, including the following:

- **Aircraft Maneuvers/Gyrations**
  - —Both Prior to and During Ejection
  - —Risk of Injury Greater at
    - · Higher G s/G Onset Rates
    - · Higher Rotational Rates and Onset Rates
- **Type Mishap**
  - —Mid-Air Collisions
  - —Hard Landing/Crash
- **Unpreparedness/Poor Body Position**
  - —For Inflight Loads
  - —For Ejection
- **Certain Types of System Malfunctions**
  - —Post Seat-Man Separation Collisions
  - —Entanglement of Man to Seat
- **Presence of Incompletely Healed Neck Injury**
- **Post-Ejection Impacts**
  - —Collisions with Trees, Buildings, Etc.
  - —Hard Landings/Poor Parachute Landing Falls

Although both the data and a qualitative analysis suggest that a relationship might exist between these factors and the incidence of "ejection associated" neck injury, it is not clear from the available data why similar incidents lead in some instances to an injury and in others fail to produce an injury. It is recognized that these data are gross field observations as opposed to controlled, measured laboratory data and, thus, lack considerable detail, especially concerning magnitude, direction, duration, and point of application of resulting forces.

In developing the above list of factors, we have examined the ejection data for relationships between the incidence of injury and the following:

- **Ejection Airspeed Mechanisms**
  - — Parachute Opening Shock
  - — Use of Ballistic Spreader Gun
  - — Lift/Drag of Helmet
- **Type Ejection Seat**
- **Ejectee Anthropometry**
- **Aircraft Maneuver/Gyration**
  - — Pre-Ejection
  - — During Ejection
- **Aircrew Restraint/Positioning/Posture**
  - — Powered versus Spring Retraction Inertia Reels
- **Canopy Mode**
  - — Jettisoned Canopy
  - — Through-the-Canopy
  - — Partial Cutting of Canopy
  - — Canopy Fragmentation
- **Existence of a Recent Prior Neck Injury**
- **Aircrew Preparedness for G Loads**

The following is a brief discussion of the above and other factors which, based upon the data and our analyses, we believe to be potentially significant to the issue of "ejection associated" neck injury.

Aircraft Maneuver/Gyration Forces

Among the potential causal factors is the aircraft inflight maneuver/gyration forces both prior to and during an ejection. Anecdotal data exists in both published [Schall] and unpublished form [Schall, and various narrative descriptions of mishaps

and inflight physiological incidents] indicating that during flight aircrew can and have incurred injuries similar in nature and severity to those reported associated with ejections.

Since 1 January 1969, reports summarized in the Naval Safety Center mishap tapes indicate that 27 U.S. Navy MA-2 restrained aircrewmen (9 during flight and 18 in other flight related events less than an uncontrolled crash) have incurred paracervical strains or sprains due soley to inflight or landing loads **(Table VII)**. The rate of such reports appears to be rising.

In addition, analyses of other types of injuries and ALSS equipment damage have clearly demonstrated the potentially dramatic effects that inflight maneuvers/gyrations can exert upon an ejecting man-seat combination as it separates from its rails. We now know, for example, that the roll rate of an aircraft can exert an enormous influence upon the behavior of such a mass. As the rate of aircraft roll during seat-aircraft separation increases, the destabilizing effects upon the man-seat combination can become very severe, resulting in rapid roll, tumble or even epicyclic tumbling of the man-seat combination and, we suspect, probably can induce ejectee injuries.

As we have earlier reported, there is a high coincidence between the reported occurrence of cervical sprains and strains and the occurrence of aircraft spins and other forms of uncontrolled flight.

There appear to be three injury mechanisms potentially inducing injuries as a consequence of the inflight aircraft maneuver/gyration forces:

Inflight

- Rapid, Inertial Load Induced Whipping Motion of the Head and Neck
- Abrupt, High Energy Contact of the Helmeted Head with Canopy or Other Object Within the Cockpit [Schall]

Ejection

- Tumbling the Seat Rapidly During Tip-Off, Inducing High Speed Separation and, Occasionally, a Subsequent Collision Between Ejectee's Helmeted Head and Seat Headrest

Bason, *et al*, demonstrated the high degree of freedom experienced by aircrew wearing their restraints properly, freedom sufficient to permit high energy helmet canopy and helmet other structure contacts induced by rapid onset negative G. The records in the pre-1969 log and the Naval Safety Center mishap tapes are replete with incidents in which such impact loadings occurred, often to be followed by an ejection as conditions worsened. In most such instances, those impacts or conditions are merely noted and not explored as a possible cause of a reported injury. Instead, the investigators' attention usually, and quite naturally, is focused upon the ejection system and environment forces as the likely culprits. And, should one desire to ascertain whether "poor position for ejection" or "windblast" or "ejection forces" or "parachute opening shock" is the likely cause, one need merely select the appropriate code(s) for a data search to obtain confirmation that such a cause is a frequently cited causal factor.

We are continuing to explore the comparative rate of occurrence of uncontrolled flight before and during ejection between the pre-1969 log data and the mishap tapes data. An initial impression is that there has been a change in the dynamic characteristics of mishaps leading to ejection with the predominant characteristics in the 1950s and early 1960s being relatively benign reliability type aircraft and aircraft systems failures. Less than a third of those early ejections appear to have involved highly dynamic uncontrolled flight, whereas nearly two thirds of ejections occurring in the 1980s appear to involve these conditions either preceding or during the ejection.

Type of Mishap

Among the 33 most severe post-1968 reported paracervical sprains and strains, 6 involved mid-air collisions (in one mishap apparently the pilot's helmet was actually impacted by the other aircraft when it penetrated his cockpit canopy and in another mishap the two aircraft collided twice!), 2 involved landing damage, one of which occurred during a ramp strike, and 16 involved spins, severe rolling and other forms of uncontrolled flight.

Unpreparedness/Poor Body Position

Again examining the more severe post-1968 paracervical sprains and strains, one was command ejected while unconscious and 4 ejected while their heads were bent forward or "snapped forward on ejection." Several less severe sprains and strains involved unconsciousness of the ejectee or other signs of unpreparedness for ejection.

In the inflight C-2 fracture and subluxation reported by Schall, the individual was unprepared when the pilot pitched the aircraft nose down. As a consequence he rose out of his seat, striking his helmet against the cockpit canopy. Subsequently, he became paralyzed when he turned his head to the right

Role of Escape System Malfunctions

Several vertebral fractures and two cervical transections occurred among aircrew:

- Experiencing post man-seat separation collisions when the empty seat overtook them following personnel parachute opening
- Who became entangled by lines tying their helmeted heads to the seat headrest prior to man-seat separation and personnel parachute deployment and opening

Recent/Existing Cervical Injuries

Three ejectees sustaining cervical sprains/strains (one pre-1969 and two in the mishap tapes) were reported to have sustained whiplash type injuries a short time preceding the flight which ended in the ejection. There were no similar reports for any of the ejectees not sustaining such injuries, although the absence of such reports might simply reflect the lack of interest resulting from the lack of an injury requiring an explanation.

Parachute Landing Falls (PLF)

Glancing collisions of descending ejectees with trees and buildings, and backward falls during landing, particularly over the ejectee's survival kit, resulting in helmeted head strikes with hard ground or pavement are relatively common. A significant proportion of those reported events are associated with ejections in which the ejectee incurred a neck injury. Again, normally such an event is not examined by the reporting medical officer as a potential cause of the neck injuries.

The majority of such incidents appear to have resulted in minor scratches or other lesser injuries. However, the potential for more serious injury appears to be present, even if overshadowed by the presence of an ejection and the more obvious forces. Accordingly, we are currently sorting and analyzing PLF versus terrain and PLF versus ejectee weight by parachute type and size data (i.e., determinants of ejectee descent velocity) to ascertain whether any pattern might exist.

Ejectee Anthropometry

Although the effects of ejectee weight might be masked by such escape system factors as the differences in the boost catapult charge, and the size, shape and porosity of the personnel parachute, no indications of a relationship between weight (lighter weight experiencing a greater boost acceleration, a lower opening shock deceleration, and a slower descent rate under a fully inflated parachute) and the incidence of neck injury have been noted within the post-1968 data. As noted previously, the pre-1969 data log does not contain any anthropometric data concerning the ejectees.

Canopy Mode

Although the effects of canopy mode upon the production of vertebral fractures and paravertebral sprains and strains has been well documented and the probable associated injury mechanism well defined, surprisingly the same pattern does not hold true for cervical fractures and paracervical sprains and strains. One factor in the failure to demonstrate the same relationship might be the small numbers of these injuries in comparison to the total numbers of the overall vertebral and paravertebral injuries. Another factor, one which seems to have reduced the overall incidence of vertebral and paravertebral injuries, is the significant reduction in the boost catapult acceleration forces exerted upon the ejectee by the more modern ejection seat systems. This lowered boost acceleration, resulting from improvements in other aspects of the systems such as personnel parachute deployment and concern over the effects of high, prolonged boost accelerations upon aircrew safety under adverse attitude escape conditions, lowers the magnitude of seat slap which can be produced by the seat-canopy impact during a through-the-canopy ejection.

However, considering the relatively high mobility of the head-neck combination in comparison to the virtually rigid immobility of the thoracic and lumbar sections of the vertebral column, it is probable that mechanisms other than seat slap in fact have been responsible for producing the cervical and paracervical injuries.

Effects of Ejection Airspeed

Many have suggested that ejection airspeed might be the critical factor, suggesting such divergent resultant effects as personnel parachute opening shock and a hangman's noose effect produced by the lift and drag characteristics of an ejectee's helmet.

It should be noted that parachute opening shock is a function of many system and parachute factors, among which are:

- Design Aspects of Ejection Seat in Which Used
  - Delay in Pack Opening
  - Type of Parachute Deployment System
  - Delay Between Pack Opening and Full Line Stretch
  - Man-Parachute Alignment at Parachute Opening
  - Orientation of the Deploying Parachute to the Airstream
- Canopy Shape (e.g. Flat Circular, Conical, Etc.)
- Canopy Size
- Porosity of the Parachute Canopy Fabric
- Control of the Canopy Throat During Deployment (to Guard Against Premature, i.e., Prior to Full Line Stretch, Inflation and Against Asymmetric Inflation of the Canopy)

and, as a consequence, when assessing the potential opening shock effects for a multitude of ejection seat systems, the effects could well mask one another. However, to avoid this masking effect, we examined the data for several individual

families of seats possessing highly common design characteristics likely to produce highly similar personnel parachute opening dynamics. We found no indications to support the idea that the "ejection associated" neck injuries might be largely produced by personnel parachute opening shocks.

The data with which to assess the potential role of aircrew helmets is at this time incomplete since we cannot ascertain the specific type of helmet employed in the majority of the ejections. We have, however, been acquiring that data and entering it into the post-1968 data base. The data have been obtained for over 950 ejectees, a number of whom sustained neck injuries, generally paracervical sprains and strains. These data, although small, have not supported the helmet hangman's noose concept.

At this stage of our investigations we have not identified any strong indications of a significant role for ejection airspeed in producing the "ejection associated" neck injuries.

CONCLUSIONS

As shown in **Table I**, serious "ejection associated" neck injuries are not a new phenomenon among U.S. Navy ejectees. What is new is the major attention now being given to this phenomenon, especially towards determining causal factors for these injuries. Much research has been conducted, is being conducted and is planned to investigate head/neck responses to dynamic conditions of impact loadings. Much of the past, current and planned investigatory research seems very narrowly focused upon dynamic responses of the head and neck to impact loadings to the exclusion of other injury mechanisms -- a few of which we have attempted to illustrate in this paper through examining and comparing the available mishap records associated with both those ejectees sustaining the serious neck injuries and those who did not.

Based upon our examinations of these mishap data we are of the opinion that there is no simple answer to the problem of "ejection associated" neck injuries; that there is no single causal factor but, rather, that the underlying causal factors are many and varied.

We are also of the opinion that the problem is an increasing one. The evidence indicates quite clearly we believe, that the frequency with which these injuries are occurring is increasing. It would also appear that cervical fracture rates are increasing, although at a slower rate than the growth in the paracervical sprains and strains. What remains to be seen following the retirement of several specific seat types, is whether the frequency of cervical transections will decline.

We are also of the opinion that a significant proportion of the serious "ejection associated" neck injuries are in fact likely to have been induced by the inflight maneuvering/gyration forces imposed upon the aircrew prior to ejection or during ejection. These, we believe, are especially significant and require consideration as helmets become the handy means for mounting sights and other needed equipment upon the aircrew. Restraint of aircrew heads in flight is probably not practical since they need exceptional head-neck and even upper torso mobility in order to visually acquire and to maintain visual contact with enemy aircraft in order to successfully engage them either defensively or offensively.

[The authors appreciate the programming and data search assistance provided by Messrs. Steve Nguyen and Charles Geiberger.]

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BIOGRAPHIES

Frederick C. Guill

Frederick C. Guill is employed in the Crew Systems Division, Naval Air Systems Command (AIR-5312C) as a Senior Engineer, Escape Systems. He attended the U.S. Coast Guard Academy, Class of '59, for 3-½ years, and graduated from the University of Washington in 1961 with a Bachelor of Science in Mechanical Engineering. In 1966, Mr. Guill earned a Masters in Engineering Administration from the George Washington University. In July 1961, he accepted employment in Crew Systems' predecessor in the Bureau of Naval Weapons. Mr. Guill was the project engineer responsible for a number of escape systems programs, including the introduction of the YANKEE type escape systems in the A-1H/J aircraft and the Stencel SIIIS-3 ejection seat into the AV-8A. He prepared the Navy's escape systems specifications for over 15 years, introducing many new technical and program management features. Currently, Mr. Guill is the Naval Air Systems Command project engineer charged with resolving the SEAWARS in-service failure problem and, also, is managing and guiding the development of the In-Service Usage Data Analysis System which he conceived and initiated. Mr. Guill has written numerous technical articles and reports concerning escape systems, ejection associated injuries, and ALSS equipment in the escape environment. His contributions to Navy aircrew safety and his accrued expertise were recognized by the SAFE Association with the award of the 1984 SAFE Award for Outstanding Contribution in the Field of Safety.

Mr. Guill is a member of the American Society of Mechanical Engineers, American Society for Metals, Human Factors Society, SAFE Association, SAE, and Aerospace Medical Association.

G. Ronald Herd

Dr. G. Ronald Herd attended the University of Kansas, earning a B.A. (1947) and an M.A. (1949), and Iowa State University, earning a Ph.D. in Mathematical Statistics (1956).

Dr. Herd has had over 30 years experience in the application of statistical and mathematical techniques to a wide spectrum of engineering problems. This experience has included applications in life testing, experimental design, quality control, and exploratory data analysis, and has covered such areas as mathematical modeling, reliability analysis, and test design for hardware systems ranging from tractors and automobiles to engines, aircraft and weapons systems.

Dr. Herd currently is president of Applied Sciences Group, Inc., and in the past has served on the Advisory Group on the Reliability of Electronic Equipment (AGREE); Bureau of Weapons Industry Maintenance Reliability Advisory Board (BIMRAD); and the U.S. Air Force Industry Advisory Committee on Weapon System Effectiveness. He participated in a review of the biological warfare R&D effort for the U.S. Army and was the Technical Director of an industry study group for the assessment of HA EMP impact on SENTINEL communications for the Army. He participated in the study of nuclear testing requirements (Project Defender). He has also served as a consultant on reliability to Centre National d'Etudes Spatiales; to the Director of Reliability and Quality Assurance, NASA, on Mercury, Gemini, Apollo, and OAO programs, and to several major industrial firms including GE, IBM, Deere, GM, RCA, as well as others.

Dr. Herd served as an Associate Editor of OPERATIONS RESEARCH from 1960 to 1970 and has published more than 35 papers in technical journals. He is a member of the Operations Research Society of America, The American Statistical Association, and Sigma Xi.

Navy Ejection Data Examined

Two Separate Populations

Unrefined State

Manually Searched Data Base	Computerized Data Base
1968 Records 1949 through 1968	2370 Records 1969 through mid-1988

Figure 1.

Navy Ejection Data Examined

Two Separate Populations

Ejection Accomplished Clear of Aircraft and within System Terrain Clearance Performance Envelope

Manually Searched Data Base	Computerized Data Base
1764 Ejection Cases	1677 Ejection Cases 1969 through mid-1988

Figure 2.

In-Envelope Ejections
(Pre-1969 Ejection Log Data)

Year	Total	Fx	Sprain/Strain
1949	1	—	—
1950	5	—	—
1951	8	—	—
1952	11	—	—
1953	29	—	—
1954	41	—	1
1955	66	—	2
1956	79	—	3
1957	114	1 (Cervical)	7
1958	134	—	12
1959	107	1 (C-6)	6
1960	104	—	9
1961	116	2 (C-2)	4
1962	105	—	8
1963	105	1 (C 1-2)	9
1964	123	—	11
1965	126	1 (Cervical Dislocation)	8
1966	126	2 (C-5 & C-2 3 4)	8
1967	176	2 (C-6*, Cervical)	7
1968	188	2 (C-4, C-7)	18
Total	1764	12	112

* Underwater

TABLE I.

In-Envelope Ejections
(Post-1968 Ejection Mishap Data)

Year	Total	Fx	Sprain/Strain
1969	215	1	20
1970	165	1	15
1971	131	6	11
1972	141	2	7
1973	114	1	9
1974	70	3	13
1975	77	—	8
1976	73	2	6
1977	82	3	8
1978	66	—	10
1979	65	2	3
1980	71	1	11
1981	61	1	12
1982	62	—	13
1983	73	1	9
1984	57	—	12
1985	35	3	9
1986	58	1	15
1987	52	—	8
1988*	5	—	—
Total	1677	28	204

* Partial Year's Data

TABLE II.

Distribution of In-Envelope Ejectees and Serious Neck Injuries by Aircraft Model and Seat Type
(Pre-1969 Ejection Log Data)

Model Aircraft/Type Seat	*Total*	*Fx*	*SPR/STR*
F2H STD	55	—	2
F7U STD	22	—	—
F9F (-2,-6) STD	102	—	8
F9F-8	40	1 (C-6)	—
F9F-8 (MkZ5)	14	—	—
F9F-8T	10	—	—
F9F-8T (MkA5)	103	1 (C-4)	5
TV	59	—	4
F3H STD	52	—	4
F3H (MkM5)	35	—	2
A4D STD	58	—	5
A4D RAPEC!	107	—	8
A4D ESCAPAC IC-3	125	1 (C 2 3 4)	18
FJ(-2,-4B) STD	112	—	5
FJ-4B (MkN5)	8	—	2
F4D STD	67	—	—
F4D (MkP5)	10	—	—
F8U INT	90	1 (Cervical)	11
F8U (MkF5)	214	2 (C-6 Cervical)	1[illegible]
F11F STD	48	—	1
F11F (MkK5)	2	—	—
T2J (LS-1)	50	2 (C-2 Dislocation Cervical)	7
A3J (HS-1)	39	1 (C-1 C-2)*	3
TT-1	1	—	—
F-4 (MkH5)	198	1 (C-5)	13
F-4 (MkH7)	12	—	0
A-6 (MkGRU5)	47	—	1
A-7 (ESCAPAC IC-2)	33	1 (C-7)	1
OV-10 (LW-3B)	2	—	—

* Aircr... 6 N observed hanging mot onless under chute ... lost

TABLE III.

Distribution of In-Envelope Ejectees and Serious Neck Injuries by Aircraft Series
(Post-1968 Ejection Mishap Data)

Series Aircraft	*Total*	*Fx*	*SPR/STR*
A-4	157	7	17
TA-4	201	5	24
RA-5C	28	3	3
A-6	140	4	13
EA-6B	80	—	13
A-7	241	6	38
TA-7	14	—	2
AV-8	38	1	10
TAV-8	3	—	1
F-4	390	4	39
F-5	3	—	—
F-8	100	—	6
F-9	7	—	—
F-14	131	1	17
F-18	15	—	5
S-3	23	—	3
T-1A	4	—	—
T-2	54	—	7
T-33	1	—	—
TF-9J	21	—	1
OV-10	26	—	5

TABLE IV.

Periods Ejection Mishap Aircraft Models Appear In Pre-1969 Log

Aircraft Model	Period Appearing in Log	Aircraft Model	Period Appearing in Log
F2H-1	1949 - 1950	F8U-1	1956 - 1968
F2H-2	1950 - 1957	F8U-2	195[illegible] - 1968
F2H-3	1953 - 1957	F-8C	1963 - 1968
F2H-4	1954 - 1959	F-8D	1963 - 1967
		F-8E	1963 - 1968
F7U-1	1950 -	F-8G	1968 - 1968
F7U-3	1954 - 1957	F-8H	1968 - 1968
		F-8J	1960 - 1968
F9F-2	1950 - 1957		
F9F-4	1952 - 1954	F4D-1	1956 - 1965
F9F-5	1951 - 1956		
F9F-6	1953 - 1962	F11F-1	1957 - 1964
F9F-7	1954 - 1956		
F9F-8	1955 - 1968	T2J-1	1959 - 1968
F9F-8T	1959 - 1968	T-2B	1967 - 1968
TV-1	1955 -	A3J-1	1959 - 1965
TV-2	1953 - 19[illegible]8	RA-5C	1964 - 1960
T-1A	1958 - 19[illegible]0		
F3H-1	1954 -	TT-1	1959 -
F3H-2	1955 - 1964		
		F4H-1	1961 - 1967
A4D-1	1954 - 1968	F4H-2	1963 - 1968
A4D-2	1958 - 1963	F-4J	1967 - 1968
A-4C	1963 - 1969	A-6A	1964 - 1968
A-4E	1963 - 1969		
A-4F	1968 - 1968	A-7A	1966 - 1968
TA-4F	1967 - 1968	A-7B	1968 - 1968
FJ-2	1955 - 1958	OV-10	1967 - 1968
FJ-3	1955 - 1963		
FJ-4	1956 - 1965		
AF-1E	1958 - 1964		

TABLE V.

Periods Ejection Mishap Aircraft Series Appear in Post-1968 Data Base

Aircraft Series	Period Appearing in Data Base
TF-9 (F9F-8T)	1969 - 1973
T-33 (TV-2)	1969 -
T-1A	1969 - 1970
A-4	1969 - 1987
TA-4	1969 - 1988
F-8	1969 - 1985
F-6	1969 - 1970
T-2	1969 - 1987
RA-5C	1969 - 1974
F-4	1969 - 1988
A-6	1969 - 1987
A-7	1969 - 198[illegible]
TA-7	1979 - 1984
OV-10	1969 - 1985
S-3	1973 - 1985
AV-8	1973 - 1988
TAV-8	1980 - 1983

TABLE VI.

MA-2 Restrained Non-Ejectees with Serious Neck Injuries
(Post-1968 Ejection Mishap Data)

Injury Sustained During Normal Flight Phases	
• Overstressed Aircraft in ACM, Rear Seat Occupant	1
• 9G Pullout, Rear Seat Occupant	1
• DACM, 7.8G	1
• 8 to 8.5G Pullout	1
• Arrested Landing	1
• Impacting Helmet Against Canopy During Arrestment	1

Injury Sustained During Flight Related Incident	
• Manual Bailout Following Uncontrolled Spin in T-34C	3
• Manual Bailout, Descending A-3 Series Aircraft	4
• Nose Gear Collapse Following Landing Aboard Carrier	1
• Landing Short of Runway, Nose Gear Collapsed	1
• Hard Landing in Overrun Area	2
• Running Off End of Runway	1
• Wheels Up Landing	2
• Ditching	2
• Inflight Engagement of #1 CDP Followed by Nose Landing Gear Collapse	1
• Engine Failure While in 80' Hover, Face Struck HUD Glass	1

TABLE VII.

Flexion, Extension and Lateral Bending Responses of the Cervical Spine

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ABSTRACT

The lateral, anterior and posterior passive bending responses of the human cervical spine were investigated using unembalmed cervical spinal elements obtained from cadavers. Bending stiffness was measured in six modes ranging from tension~extension through compression~flexion. A five-axis load cell was used to establish the end conditions. Results include moment-angle curves, relaxation moduli and the effect of cyclic conditioning on bending stiffness. The Hybrid III ATD neck was also tested and its responses are compared with the human. It was observed that the Hybrid III neck was more rate sensitive than the human, that mechanical conditioning significantly changed the stiffness of the human specimens and that changing the end condition from pinned-pinned to fixed-pinned increased the stiffness by a large factor. The bending stiffness was significantly influenced by the direction of the bending moment, the type of end restraint, the magnitude of the deformation and the previous deformation history. The shear force produced by the end conditions was an important factor in the applied moment. This shear force not only changes the moment acting on the specimen but also influences the failure mode. These experiments indicate that when the loading is eccentric (as it almost always is), the primary deformation mode is bending, and the moment applied to the specimen is strongly influenced by shear forces and the magnitude of the eccentricity. The axial load is therefore a poor indicator of the type and magnitude of failure stresses.

MR and CT was used to visualize the damage after loading. When compared to the dissection results MR was clearly superior to CT in detecting soft tissue and ligamentous injuries.

INTRODUCTION

The majority of the studies of the structural properties of the spine have involved compression. Perhaps the earliest such study was Messerer's work on the mechanical properties of the vertebrae (2). He reported compression breaking loads ranging from 1.47-2.16 kN for the lower cervical spine. Bauze and Ardran loaded human cadaveric cervical spines in compression and reported forward dislocations with loads of 1.32 1.42 kN (3). However, their experiments were designed to force the dislocations to occur at a given vertebral level. Sances tested isolated cadaver cervical spines in compression, tension and shear (4). A quasi-static compression failure was observed at a load of 0.645 kN, and dynamic compression-flexion failures were reported at loads ranging from 1.78-4.45 kN. McElhaney *et al.* applied time-varying compressive loading to unembalmed human cervical spines (5,6). Failures were produced which are similar to those observed clinically with maximum loads ranging from 1.93-6.84 kN. In addition, it was found that small eccentricities in the load axis could change the buckling mode from posterior to anterior. Panjabi *et al.* measured rotation and translation of the upper vertebra as a function of transection of the components in single units of the cervical spine (7). Selecki and Williams conducted a study of cadaveric cervical spines loaded with a manually operated hydraulic jack (8). They were able to duplicate several types of clinically observed injuries, but reported loads in terms of the hydraulic pressure. Nusholtz *et al.* studied neck motions and failure mechanisms on unembalmed cadavers due to crown impacts; failure loads ranged from 3.2 to 10.8 kN (9). They reported that spinal response and damage were significantly influenced by the initial configuration of the spine.

Very few tests have been conducted on longer spinal segments. Edwards *et al.* tested lumbar spine motion units in combined loading (10). They found that stiffness of the motion unit was nonlinear and increased with increasing load. Markolf and Steidel tested human cadaveric thoracolumbar spine motion units in flexion, extension, lateral bending, torsion, and tension (11). They conducted free vibration tests and reported stiffness and damping values for the various test modes and vertebral levels. Panjabi *et al.* measured the three-dimensional stiffness matrix for all levels of the thoracic spine by measuring all components of deflection of spinal units for various loading modes (12). Roaf loaded single cervical spinal units in compression, extension, flexion, horizontal shear, and rotation (torsion) (13). He found that the intact disc, which failed at approximately 7.14 kN, was more resistant to compression than wet vertebrae which failed at approximately 6.23 kN. It is his contention that ligamentous rupture cannot be caused by hyperflexion or hyperextension, but only by rotation and/or shear forces. Tencer *et al.* performed static tests on individual lumbar spinal units (14). They presented load-deflection data for all loading modes. Hodgson measured the strain at selected locations of the cervical vertebrae of cadavers under several head impact modes (15). He concluded that the effects of off-axis, torsional and transverse shear are important variables and influence the axial response. Seemann compared the dynamic responses of the human and Hybrid III neck (16). He concluded that there was a good match with some bending modes but a poor one in others. An extensive review of the literature was presented by Sances 1981 (1).

A major problem with tests on spinal elements has been the proper measurement of the forces and moments applied to the specimen. The experiments reported here used a five-axis load cell in an attempt to better understand the reasons for the wide range of compressive failure loads and failure mechanisms reported in the literature.

METHODS

SPECIMEN TYPES AND PROCUREMENT - Unembalmed human cervical spines were obtained shortly after death, sprayed with calcium buffered, isotonic saline, sealed in plastic bags, frozen and stored at -20°C. Cervical spine specimens generally included the base of the skull, approximately two centimeters around the foramen, or C1 at the superior end and C5, C6, C7, or T1 at the inferior end. The associated ligamentous structures were kept intact. X-rays were taken and reviewed to assess specimen integrity. Medical records of donors were examined to ensure that the specimens were normal for their age group and did not show evidence of serious degeneration, spinal disease, or other health-related problems that would affect their structural responses.

SPECIMEN PREPARATION - Prior to testing, each specimen was thawed at 20°C for 12 hours. The pre-test specimen preparation was performed in an environmental chamber, which was designed to prevent specimen dehydration and deterioration. A variable flow humidifier pumped water vapor into the chamber to create a 100% humidity environment. The end vertebrae were cleaned, dried, and defatted for casting. The specimen was mounted in aluminum cups with a pin inserted into the spinal canal in order to provide a reference bending axis. Using polyester resin, the ends of the specimens were cast in the cups so that the cups were approximately perpendicular to the axes of the end vertebrae (17). During casting, the aluminum cups were cooled in a flowing water bath to minimize degradation due to the heat of polymerization.

TEST INSTRUMENTATION - A Minneapolis Test Systems (MTS) servo-controlled hydraulic testing machine was used to conduct the various viscoelastic tests. The first series was axial compression using a spherical washer to minimize the moments at the ends. A lead screw adjustment at the lower end was used to straighten the lordotic curve and align the specimen (Figure 1).

The second series was a combination of bending and axial loading. An eight-channel transducing system was used to measure the axial, lateral, and anterior forces, the flexion-extension and lateral bending moments, the linear motion of the ram, and the angular motion of the specimen ends. Loads and moments were measured with a five-axis load cell assembly, which was constructed using two GSE three-axis ATD neck load cells. The motion of the specimen ends was measured with an internal coaxial linear variable differential transformer (LVDT) and two external rotational variable differential transformers (RVDT). These transducers provided data to establish the motion of the two specimen ends from direct measurements of the total bending angle and calculations of the specimen length change. The internal LVDT was used to monitor the ram motion and hence the displacement of the clevis end of the lower transfer bar. One external RVDT was used in the pinned-pinned and fixed-pinned tests to track the rotation of the specimen end of the lower transfer bar relative to the ram; the second external RVDT was used in the pinned-pinned tests to track the rotation of the specimen end of the upper transfer bar. Figure 2 is a schematic diagram of the test apparatus.

A digital measurement and analysis system was developed utilizing a data logging computer. The multichannel microcomputer-based data acquisition system incorporated an RC Electronics ISC-16 Computerscope for the digitization and storage of data. This system, which consists of a 16-channel A/D board, external instrument interface box, and

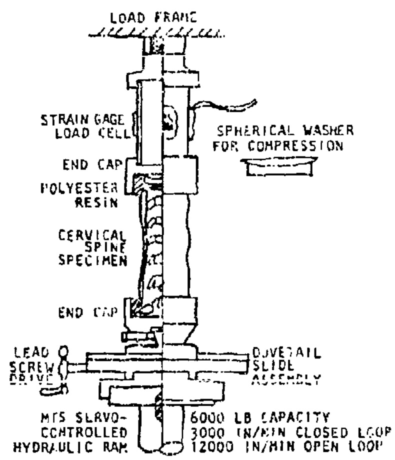

Figure 1. Axial Compression Test Fixture.

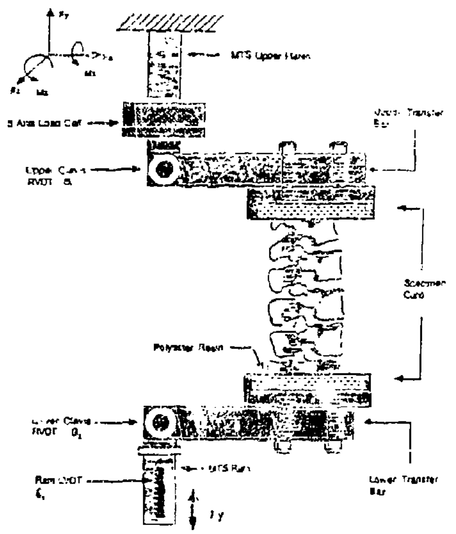

Figure 2. Free-Free Test Configuration

Scope Driver software, has a 1 MHz aggregate sampling rate capability with 12 bit resolution and writes data directly to a hard disk. In addition, during the failure tests, flouroscopic images were recorded on videotape.

THE COMBINED AXIAL LOADING - BENDING TEST APPARATUS – A specially designed test jig was developed to place the specimen in a state of eccentric axial loading. This resulted in a combined axial load and bending moment applied at the ends of the specimen. The apparatus provided adjustable moment arms and accommodated the following six test modes: compression-flexion (CF), tension-flexion (TF), compression-extension (CE), tension-extension (TE), compression-lateral bending (CL), and tension-lateral bending (TL). Two test configurations were utilized: (1) pinned-pinned end conditions (PP), and (2) fixed-pinned end conditions (FP).

For the pinned-pinned end conditions, the upper transfer bar was attached via a clevis to the load cell assembly, which was rigidly mounted to the upper platen of the MTS. The lower transfer bar was attached via a clevis to the ram of the MTS. The centerline of the specimen was parallel to, but not coincident with, the line of action of the MTS ram. The clevis end of the upper transfer bar was constrained from translation. The two external RVDTs were mounted on the test apparatus in order to measure the angular displacement of each transfer arm. In this configuration, the specimen was mounted with the superior end attached to the upper transfer bar and the inferior end attached to the lower transfer bar.

For the fixed-pinned end conditions, the upper clevis and corresponding RVDT were removed. In this configuration, the specimen was mounted with the superior end attached to the pivoting lower transfer bar and the inferior end fixed to the load cell assembly, which was rigidly mounted to the upper platen of the MTS.

A free body diagram of the test configuration is presented in Figure 3. The reference center line of the specimen is the central axis of the spinal foramen. The moment at the center of the specimen is

$$M_A = P_y a - P_x b,$$

and the moment measured by the load cell is

$$M_0 = F_x B.$$

The moment induced by the shear force P_x was significant in the fixed-pinned configuration but was negligible in the pinned-pinned configuration. The apparatus had minimal overshoot and vibration below test frequencies of 5 Hz. Inertial forces begin to predominate above 10 Hz, and this is the current system's upper frequency range.

In this paper, test rates will be described in Hertz. The test period is the reciprocal of the frequency, and the time to peak load is one-half of the test period. The deformation rate is the maximum deformation in angular or linear units multiplied by twice the test frequency.

CONSTANT VELOCITY TESTS - Constant velocity tests were conducted on mechanically stabilized spines using triangle wave deformations at frequencies of 0.01, 0.1, 1.0, 5 Hz, and, for some specimens, 10 Hz. Thus, the deformation rate was varied by a factor of 500-1000.

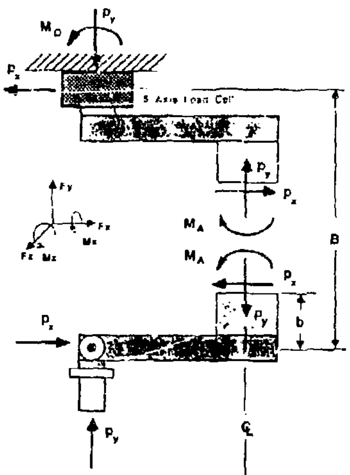

Figure 3. Freebody Diagram for the Fixed-Pinned Test Configuration.

Typical constant velocity moment-angle curves are presented for human and Hybrid III cervical spines in the pinned-pinned and fixed-pinned test configuration in Figures 4 and 5. All of the curves exhibit a hardening response (increasing stiffness) and hysteresis. The human and Hybrid III responses are fundamentally different. The Hybrid III shows the classic linear viscoelastic response of increasing stiffness with displacement rate while the human shows little change in stiffness or hysteresis over the rate range tested. Since these features of hysteresis, relaxation, and stiffness are not very sensitive to the rate of strain, simple linear viscoelastic models would not be appropriate predictors of the time dependent human spinal bending responses; and the more complex Maxwell-Weichert quasi-linear model is required (6).

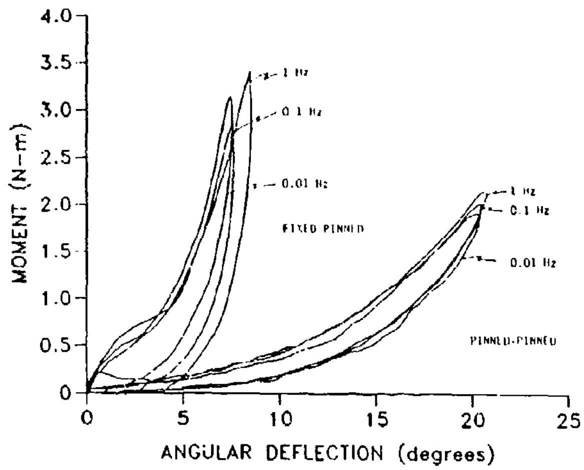

Figure 4. Typical Constant Velocity Profile for Human Cervical Spine (Compression-Flexion).

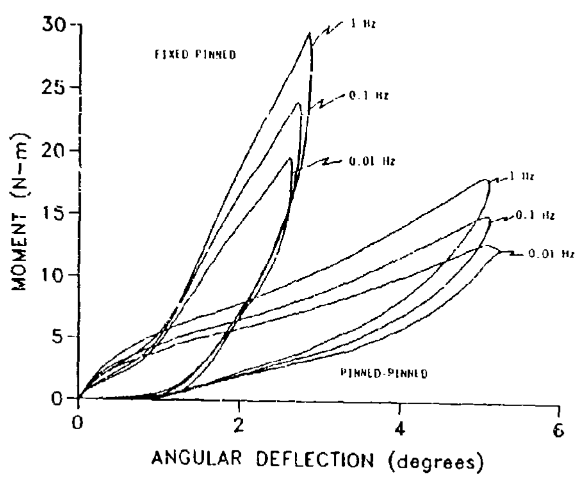

Figure 5. Typical Constant Velocity Profile for Hybrid III Neckform (Compression-Flexion).

TABLE 1. CONSTANT VELOCITY STIFFNESS (N-m/rad).

MODES	HUMAN						HYBRID III	
	FIXED-PINNED			PINNED-PINNED			FIXED-PINNED	PINNED-PINNED
	Mean	σ	N	Mean	σ	N	Mean	Mean
CF	29.9	2.6	10	8.1	0.7	5	589.1	150.8
TF	41.8	5.6	5	14.8	1.3	5	608.4	199.0
CE				2.8	0.6	9	795.7	122.5
TE	309.0	26.9	5	10.3	1.2	11	232.1	138.8
CL	8.7	0.6	10	3.1	1.0	17	898.9	190.9
TL	254.1	34.6	5	13.0	1.9	5	442.0	226.1

σ = Standard Deviation; N = Number of Tests.

Table 1 shows the stiffness averaged over four rates for all specimens. Three distinct tests of the Hybrid III were performed so that each value represents the mean of 12 tests. Several observations are apparent from this data. First, there are significant differences between the bending stiffnesses of the cadaver cervical spine and the Hybrid III. Second, there are significant differences in the bending stiffness of the cadaver cervical spine in the different modes. Tension-extension was the largest with a stiffness of 125 N-m/Radian, fixed-pinned and 15 N-m/Radian, pinned-pinned. Compression-lateral was the smallest with a stiffness of 10 N-m/Radian, fixed-pinned and 2.6 N-m/Radian pinned-pinned.

Figure 6 shows a typical response pattern for the human cervical spine to the various combined bending and axial loading modes. Figure 7 shows a typical response pattern for the Hybrid III.

Constant velocity testing in axial compression was also performed on fourteen specimens The average stiffness per motion segment was 571 newtons per centimeter. Typical test results for a single motion segment are shown in Figure 8.

FAILURE TESTS - After the battery of viscoelastic tests was accomplished, a constant velocity failure test at 0.1 Hz was performed on the bending test specimens. This rate was used so that flouroscopic images of the specimen motion could be obtained. All failure tests were in the compression-flexion mode (CF). After the tests the specimens were examined with magnetic resonance imaging (MRI) and computerized tomographic radiography (CT), then dissected. Table 2 provides the maximum moment axial force and shear force applied to the specimen and the bending angle at which these peaks occurred. The first four tests (1C, 2C, 3C, 4C) were performed in the pinned-pinned mode and the remainder (6C, 7C) were tested in the fixed-pinned mode. In the pinned-pinned configuration the specimens were very flexible and were able to bend through on average of 45 degrees without an unstable dislocation. These specimens contained C_1 through T_1 and seven intact intervertebral structures. This is approximately 6.4 degrees per vertebral level. The shear forces were very small. The axial forces were low enough that the major stresses were due to the bending moment. The primary failure mechanism was disruption of the interspinous ligaments (ligamentum nuchae), the ligamentum flavum and capsular ligaments. There was also minor anterior wedging of the middle vertebral bodies and discs. In the pinned-pinned configuration the moment is maximum in the middle of the specimen. This may be the reason that the most frequent spinal cord injury level observed clinically is C_4 - C_5 and C_5 - C_6 (5).

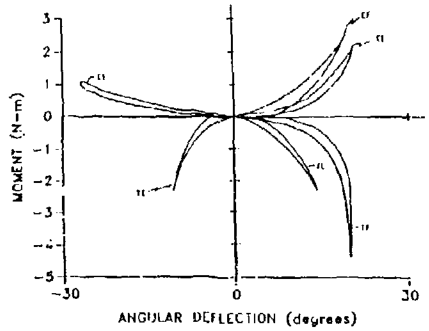

Figure 6. Typical Bending Responses of Human Cervical Spine.

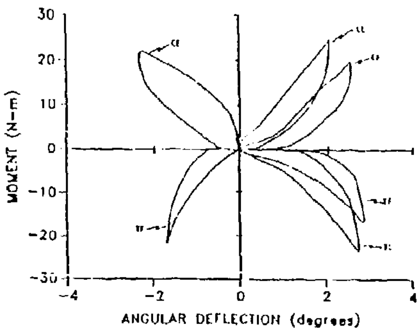

Figure 7. Typical Bending Responses of Hybrid III Neckform.

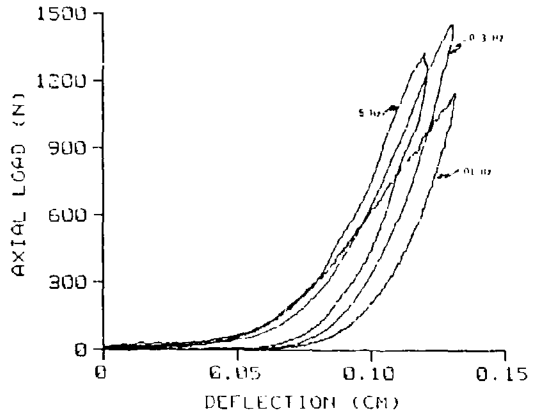

Figure 8. Typical Constant Velocity Axial Load Response C5-C6 Motion Segment.

TABLE 2. FAILURE TEST RESULTS.

SPECIMEN NUMBER	AGE/SEX	VERTEBRAL LEVELS	MAXIMUM MOMENT (N-m)	MAXIMUM AXIAL FORCE (N)	MAXIMUM A-P SHEAR (N)	ANGLE AT MAX.MOMENT (deg)	FAILURE CLASSIFICATION
1C P-P	52/M	$C_1 - T_1$	14.6	192	0	54	C_4-C_5, C_5-C_6 ligamentum nuchae, ligamentum flavum, and post. long. ligament torn
2C P-P	64/F	$C_1 - T_1$	8.75	214	0	57	C_6-C_7 ligamentum nuchae and R capsular ligament torn
3C P-P	N/A	$C_1 - T_1$	3.01	108	0	31	wedging of C_4-C_5 bodies, C_5-C_6 ligamentum nuchae disrupted
4C P-P	69/M	$C_1 - T_1$	3.40	338	11.7	40	wedging and broadening of C_4-C_5 and C_5-C_6 bodies, tear of C_3-C_4 disc
5C P-F	77/M	$C_1 - T_1$					this specimen was not loaded to failure
6C P-F	76/M	$BOS - T_1$	6.7	1513	23.0	15	C_4-C_5 ant. disc disrupted, C_2-C_3, C_3-C_4, C_4-C_5 L capsular ligaments partially disrupted
7C P-F	66/M	$BOS - T_1$	10.2	2305	35	22	C_4-C_5, C_5-C_6, C_6-C_7 shortened discs and wedged bodies, disrupted C_7-T_1 disc, ligamentum nuchae and ligamentum flavum stretched

In the fixed-pinned configuration much larger axial forces are required to produce the same bending moment because the shear force produces a counteracting moment. This is reflected in the failure mechanisms by superimposing compressively induced failures (wedging of bodies and discs) to the posterior tensile failures due to bending.

Figure 9 shows a composite of the moment angle diagrams for the failure tests. The maximum moment ranged from 3.01 to 14.6 N-m. This large range is probably due to the variation in the size of the specimens. Specimen 1C and 7C had much larger vertebrae than the others as demonstrated by the CT scans.

In the axial compression mode the failure test was performed at a ram velocity of 64 cm/sec.

Table 3 summarizes the type of failure, the maximum load and deflections, and the strain energy or area under the loading portion of the load–deflection curve failure. Figure 10 shows a representative curve.

The following four failure mechanisms were observed for the axial compression tests.

EXTENSION/COMPRESSION - As the body, discs and facet joints resisted the load, the posterior elements were compressed and, as failure of the disc and end plates occurred, the cervical spine extended in a forward buckling mode. Specimen A80-339 failed in this way with rupture of the anterior longitudinal ligament and distraction of the anterior section of the disc between C4 and C5. This occurred with a one centimeter posterior eccentricity.

JEFFERSON FRACTURES - In the clinical literature, the common etiology of a fracture of the atlas is a direct blow to the top of the head. In these tests, the experimentally produced atlas fractures, which were usually bilateral and symmetrical, involved the anterior and posterior arches. This was probably due to the compressive force driving the articular condyles outward and bending the arches.

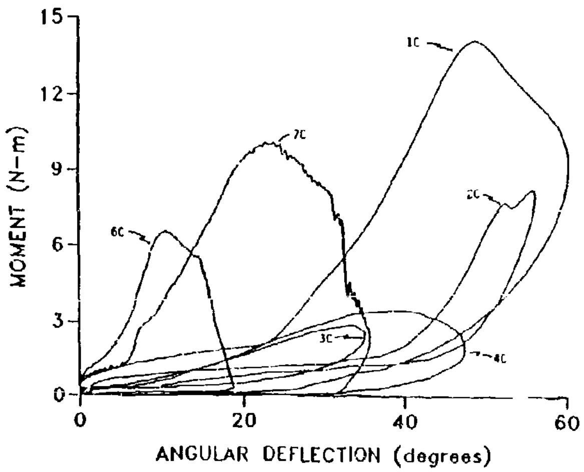

Figure 9. Failure Curves -- Compression-Flexion.

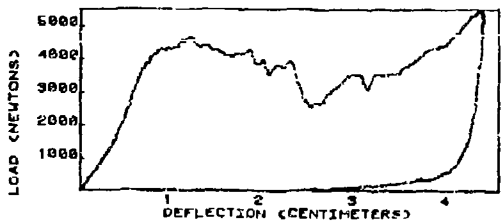

Figure 10. Typical Axial Compression Test (A83-26).

TABLE 3. AXIAL COMPRESSION FAILURE TESTS.

Specimen Number	Age (years) Sex	Description	Failure Mode	C5 Area (cm^2)	Max. Load (N)	Max. Deflection (cm)	Strain Energy (N-cm)
A79-409	58M	B.O.S.* to T2	Jefferson Fr.	5.71	3560	3.0	7470
A79-415	37M	B.O.S. to T1	Compression C5	5.98	5340	3.0	12800
A79-419	49F	B.O.S. to T2	Compression C4&C5	4.29	4860	3.0	10300
A79-423	52M	B.O.S. to T1	Jefferson Fr.	6.17	4190	3.0	7920
A79-431	46M	B.O.S. to T1	Anterior Wedge C5	6.30	4720	3.0	9340
A80-289	70M	B.O.S. to C7	C2 Cracked	5.43	5010	2.9	7950
Retest		C3 to C7	Anterior Wedge C6		6040	2.7	10900
A80-339	62F	B.O.S. to T1	Extension Failure	3.51	1930	4.0	4480
A80-352	62M	B.O.S. to C6	Jefferson Fr.	6.58	3120	3.0	5740
A80-357	46F	B.O.S. to C6	Jefferson Fr.	3.71	960	2.9	1800
A80-364	41M	B.O.S. to C6	C1&C2 Fractured	5.62	5270	2.5	8550
A80-368	77M	B.O.S. to C6 C3,4,5 Bodies Fused	C1 Fractured	5.77	3650	2.7	6350
A80-384	64F	B.O.S. to C7	C2 Fractured	4.38	4060	4.5	12300
Retest		C3 to C7	Burst C4 and Anterior Wedge C4&C5		6840	3.5	15500
A83-26	44M	C2 to T2	Burst Fracture C3, C4&C5	5.45	5470	4.4	15600
A83-42	63F	B.O.S. to C6	Burst Fracture C3&C6	3.28	3000	2.8	5550

*B.O.S. = Base of Skull

BURST FRACTURES – Comminuted vertical fractures through the vertebral body produced framentation of the centrum into a number of large pieces. There were no obvious areas of compressed cancellous bone. Analysis of x-rays taken before and after each test indicated that the specimens that burst were slightly flexed to straight while the specimens that sustained the Jefferson fractures were slightly extended to straight. The burst fracture required larger forces and strain energies than the Jefferson fractures. The load~deflection diagram exhibited a characteristic M-shape or twin peak. Specimen A80-384 showed multiple spikes in the first peak which may be related to the multiple fracturing process.

ANTERIOR WEDGING – The addition of the small flexing moment arm (h < 1 cm) using the test fixture resulted in compression and fracture of the anterior section of the vertebral body. The addition of a slightly larger moment arm (h = 1 cm) produced buckling rearward. Pieces of the cortical shell were displaced in a random pattern. End plate failure occurred and the intervertebral disc was disrupted. However, the amount of displacement applied to the specimen did not result in large anterior dislocation or rupture of the anterior longitudinal ligament. By careful alignment and adjustment of the slide-positioning device, we were able to produce fractures similar to those observed clinically. But, after fourteen tests, we had the distinct impression that one or two millimeters forward or backward, right or left, made a tremendous difference in the outcome. Perhaps, this is the reason there is such a wide range of responses to cervical spine compression in the relevant literature.

SUMMARY – In the engineering disciplines, a designer starts with a basic building material and shapes it into a structure with specified load and deformation responses. These load and deformation responses are defined as the structural properties. The structural properties are determined by the size, shape, configuration and material of which a structure is composed. In contrast, the material properties are independent of the structure or shape of the material under consideration. Since the human body exists, it exhibits load and deformation responses which determine its injury potential in traumatic environments. Knowledge of the properties of the material of which the human body is composed is useful in so far as it leads to a better understanding of these structural properties.

This study demonstrated the complex, time-dependent responses of the human cervical spine and the Hybrid III neckform in combined axial and bending deformations. In all test modes (axial compression, tension-extension, tension-flexion, tension-lateral bending, compression-extension, compression-flexion, compression-lateral bending) there was a large difference between the responses of spines in the fully equilibrated and mechanically stabilized states. In all test modes, the time-dependent responses included a significant viscoelastic exponential relaxation. The hysteresis and stiffness of the human specimens was only weakly dependent on strain rate.

There was a significant difference between the stiffness of the cadaver cervical spines and the Hybrid III. This was expected, since the performance requirements of the Hybrid III were based on human volunteer data, and it is considered to represent a tensed human neck while the cadaver spines have no musculature present (19). The Hybrid III responses were the typical linear viscoelastic type. That is, a linear differential equation would provide an adequate model. The behavior of the human cervical spine was more complex, however, and requires a quasi-linear model (6).

The bending stiffness of the cervical spine was significantly influenced by the direction of the bending moment, the types of end restraint, the magnitude of the deformation, and the previous deformation history. After approximately thirty deformation cycles a mechanically stabilized state was attained that provided repeatable load-deformation responses. The tensile modes were consistently stiffer than the compressive modes. This may be due to a shift in the neutral axis toward the tensile side which pre-tensions slack ligaments and reduces the eccentricity.

Simple beam theory predicts doubling of the bending stiffness when comparing pinned-pinned and fixed-pinned ends. These tests showed an increase in stiffness of approximately eight times. The test apparatus used in these tests (and by most other researchers) constrained the pinned end to move in a straight line. This produced a shearing force which, acting over a relatively long moment arm, stiffened the specimen. This shearing force not only changes the moment acting on the specimen but also influences the failure mode. Several researchers have tested cervical specimens without well controlled and monitored end conditions. Most other works report only the axial load. These experiments indicate that when the loading is eccentric (as it almost always is), the primary deformation mode is bending; and the moment applied to the specimen is strongly influenced by shear forces and the magnitude of the eccentricity. The axial load is therefore a poor indicator of the type and magnitude of failure stresses.

After failure loading many of the specimens imaged with plain radiographs, computed tomography and 1.5 Tesla MRI to detect patterns of injury and to determine the efficacy of each imaging modality in detecting spinal injury.

Complete tears, buckling and stripping, as well as more subtle disruptions of the ligamentum flavum, capsular, anterior and posterior longitudinal ligaments were observed on MR examination. Over 90% of the ligamentous injuries were accurately depicted by MR. MR was clearly superior to CT in detecting soft tissue and ligamentous injuries. Studies in patients suggest that MR demonstration of these injuries *in vivo* is also feasible.

ACKNOWLEDGEMENTS

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A Kinematic/Dynamic Model for Prediction of Neck Injury during Impact Acceleration[1]

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SUMMARY

A statistical study was made of six head kinematic response curves for a set of 57 human and 29 animal (rhesus) -X impact acceleration tests conducted at the Naval Biodynamics Laboratory. The acceleration levels ranged from six to fifteen g's for humans and 42 to 106 g's for animals. The six analyzed responses included the X and Z components of the linear acceleration and displacement and the Y axis angular acceleration and displacement. Each head kinematic response variable was non-linearly regressed on sled acceleration profile and head orientation parameters. Regression equations for rhesus and human kinematics had the same exponential functional form with correlations ranging from 0.50 to 0.95. Statistical measures of goodness-of-fit were highly significant.

The results confirm that the rhesus head/neck is a good biomechanical model for the human. Extension of this approach can lead to the scaling of kinematics between humans and animals which can be used to develop an injury prediction model for humans. Future applications include re-analysis of previous results on the effects of mass distribution parameters on head/neck dynamic response.

INTRODUCTION

Aircrew injuries sustained during emergency egress and recovery are a ubiquitous source of loss in military aviation. The Naval Biodynamics Laboratory (NAVBIODYNLAB) is studying human[2] and rhesus monkey[3] head and neck response to whole-body acceleration to develop predictive models for neck injury. These models can also be used to predict the effects of added head mass and shifts in head center-of-gravity as produced by head-mounted systems. While human head-neck kinematics for the -X vector direction have been successfully modelled utilizing a deterministic head-neck linkage model [1,6-10] driven by accelerations at T-1, the first thoracic vertebral body, the lack of rhesus T-1 data precludes the development of a similar animal model. The large database of rhesus kinematic, injury and pre-injury data [11,12] collected at the NAVBIODYNLAB requires other means for scaling human and rhesus head kinematic responses. Although no deterministic linkage model is available for the rhesus kinematics, preliminary results [13] provided evidence that, except for scale, the underlying structure of key human and rhesus responses is essentially the same. This paper extends those results and provides a regression model for human and animal kinematics using sled acceleration profiles and head initial orientation parameters.

METHODOLOGY

(1) Database. The data used in this analysis were a subset of the large amount of human and rhesus kinematic -Gx data collected over the years at NAVBIODYNLAB. The experimental and instrumentation details have been extensively reported elsewhere [2-5, 11]. Briefly, the human volunteers and animals are instrumented to measure head displacement and linear and angular acceleration. The subjects are seated with full torso restraint and the unencumbered head and neck are allowed to move freely. Table 1 contains the parametric details for the 57 human sled acceleration profiles and Table 2 the details for the 29 animal profiles. The identified parameters include peak sled acceleration (PSA), rate of acceleration onset (ROO), endstroke sled velocity (ESV), and the duration of peak acceleration (DOP). Figure 1 illustrates the time trace of a human and animal sled acceleration. The six kinematic variables studied were the head X and Z linear displacement (DAX, DAZ respectively) and acceleration (AAX, AAZ) and Y angular displacement (PHB) and acceleration (QHB). DAX, DAZ, AAX, and AAZ measured the kinematics of the head anatomical origin with respect to the sled coordinate system. PHB and QHB were measured with respect to the Y-axis of the head anatomical coordinate system. An example of an original set of data is shown in Figure 2 which is a plot

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[1] The interpretations and opinions in this work are the author's and do not necessarily reflect the policy and views of the Navy or other government agencies.

[2] Volunteer subjects were recruited, evaluated, and employed in accordance with the procedures specified in the Department of Defense Directive 3216.2 and Secretary of the Navy Instruction 3900.39 series. These instructions meet or exceed prevailing national and international standards for the protection of human subjects.

[3] The animals used in this work were handled in accordance with the principles outlined in the guide for the care and use of laboratory animals (National Institutes of Health Document No. NI480-23) established by the Institute of Laboratory Animal Resources, National Research Council, Bethesda, MD.

of the human X axis linear head displacements (DAX). Figure 3 is a similar plot of the set of animal DAX data.

Table I

HUMAN SLED ACCELERATION PARAMETERS

RUN	PSA (m/s^2)	ROO (m/s^3)	ESV (m/s)	DOP (ms)
LX3856	61	1370	10.0	118.5
LX3870	61	1356	10.0	119.4
LX3872	60	1344	9.9	121.5
LX3876	59	1328	9.8	119.8
LX3878	61	1395	10.0	118.6
LX3880	61	1328	10.0	118.7
LX3883	80	2023	12.0	113.0
LX3886	80	1990	12.0	114.6
LX3887	59	1315	9.8	120.2
LX3890	80	2007	12.0	113.2
LX3894	82	2008	12.1	111.0
LX3895	81	2020	12.0	112.2
LX3898	82	2023	12.1	114.4
LX3901	78	1955	11.9	116.5
LX3903	100	2781	13.7	106.7
LX3914	100	2675	13.8	105.3
LX3916	101	2705	13.8	104.9
LX3918	100	2870	14.0	109.3
LX3920	121	3837	15.4	101.7
LX3921	118	3744	15.1	100.8
LX3924	122	3791	15.4	98.5
LX3926	119	3718	15.0	100.8
LX3927	120	3696	15.3	101.2
LX3928	101	2727	13.8	105.5
LX3939	121	3737	15.4	99.7
LX3940	121	3714	15.4	101.2
LX3941	122	3740	15.5	97.7
LX3942	118	3741	15.3	102.2
LX3946	133	4261	16.2	93.5
LX3948	134	4419	16.1	91.0
LX3949	133	4359	16.1	92.8
LX3953	130	4338	16.0	97.2
LX3954	138	4688	16.5	92.9
LX3955	133	4345	16.2	93.7
LX3957	143	4981	16.8	90.9
LX3958	143	4845	16.9	91.0
LX3959	145	4573	16.8	88.9
LX3962	138	4646	16.6	93.9
LX3963	142	4664	16.7	91.1
LX3965	143	4830	16.7	89.9
LX3968	140	4793	16.5	92.1
LX3969	151	5356	17.2	88.1
LX3970	152	5235	17.3	87.6
LX3972	151	5421	17.3	88.9
LX3982	153	5305	17.5	88.4
LX3983	153	5276	17.5	89.1
LX3985	100	2858	13.8	106.4
LX3986	152	5266	17.3	87.1
LX3987	142	4702	16.8	91.7
LX398[illegible]	100	2842	13.8	106.5
LX399[illegible]	151	5163	17.3	88.9
LX3991	100	2869	13.9	107.4
LX3993	100	2738	13.8	107.1
LX3995	100	2757	13.8	106.4
LX3997	80	1992	12.1	114.5
LX3998	100	2796	13.7	106.2
LX3999	101	2811	13.9	105.6

Table II

ANIMAL SLED ACCELERATION PARAMETERS

RUN	PSA (m/s^2)	ROO (x1000)	ESV (m/s)	DOP (ms)
LX4790	834	80	26.1	20.9
LX4791	988	113	28.3	19.3
LX4799	1040	133	29.0	19.2
LX4801	995	134	28.4	20.4
LX4803	844	99	26.4	22.4
LX4810	545	43	21.4	25.9
LX4814	692	65	23.9	23.9
LX4820	545	42	21.3	26.0
LX4822	843	86	26.2	22.1
LX5135	409	21	18.8	31.1
LX5147	728	60	24.8	22.5
LX5150	411	21	18.9	31.0
LX5153	412	20	18.9	30.5
LX5156	415	26	19.0	31.8
LX5157	411	20	18.8	28.0
LX5164	732	56	24.8	22.2
LX5165	744	71	25.3	22.8
LX5768	570	38	22.0	25.9
LX5770	559	[illegible]	22.0	27.8
LX5772	556	40	21.9	27.6
LX5774	555	52	22.0	29.6
LX5777	730	54	25.1	22.3
LX5779	554	43	21.9	28.5
LX5782	733	55	25.1	22.3
LX5784	730	53	25.0	22.6
LX5786	870	76	27.1	20.5
LX5793	880	79	27.2	20.5
LX5795	897	75	27.5	19.9
LX5797	889	83	27.5	20.8

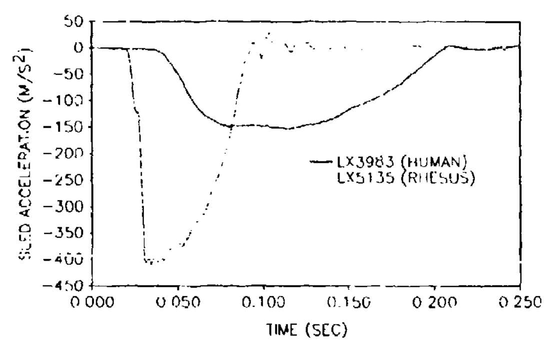

Figure 1. Comparison of sled accelerations.

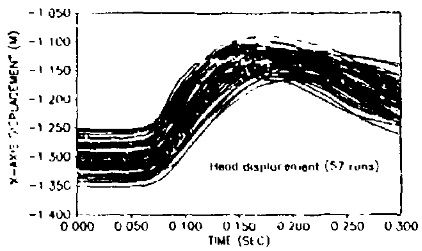

Figure 2: Human X axis linear head displacements (DAX) in the sled coordinate system.

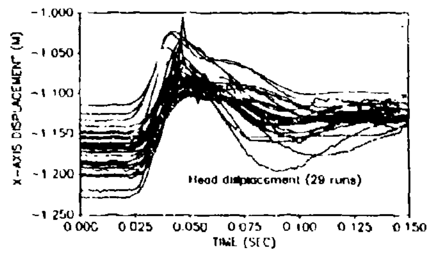

Figure 3: Rhesus X axis linear head displacements (DAX) in the sled coordinate system.

(2) Analysis. The first step in the statistical curve fitting procedure was to identify the key independent variables which form the basis of the regression procedure. The five variables previously identified [10] were used: the initial linear displacements of the head in the X (DAX_1) and Z (DAZ_1) directions, the initial rotation of the head about the head anatomical Y-axis (PHB_1), the peak sled acceleration (PSA) and the endstroke sled velocity (ESV). For the non-linear regression computations, the BMDP P3R[4] program was used to determine the parameters of an exponential model. For each of the three sets of head displacement data, two related parametric models were developed. In the first model, each displacement curve D(t) was assumed to be of the form:

$$D(t) = p_1 t^{p_2} e^{-p_3 t} \qquad (1)$$

where t is time and p_1, p_2, and p_3 are the unknown parameters estimated by the non-linear regression procedure.

These three parameters were then regressed against the five independent variables using the BMDP P6R and P9R[4] programs. The results of this regression then further improved by a changing the parameters of equation (1). This resulted in the final head displacement model of the form:

$$D(t) = q_1[(1/q_2)te^{(1-t/q_2)}]^{q_3} \qquad (2)$$

where t is time, $q_1 = p_1(q_2/e)^{p_2}$ = the peak (maximum) value of D(t), $q_2 = p_2/p_3$ = the time to the peak of the displacement curve, D(t) and $q_3 = p_2$. Equation (2) facilitates the study of the effects of the independent variables on the timing and magnitude of maximum head displacement. Confidence ranges for (2) were also computed.

RESULTS

The five independent variables were sufficient to predict all chosen head displacement data with R^2 values ranging from 0.50 to 0.95. The actual coefficients for equation (1) (and by computation, for equation (2)) are functions of the independent variables DAX_1, DAZ_1, PHB_1, PSA and ESV. Appendix I lists the matrix expressions for evaluating these coefficients for the three human and rhesus head displacement variables. Figures 4 – 9 illustrate the confidence bands for the regression results. Each figure shows the mean predicted displacement curve and the 95% confidence band. Superimposed on each plot is a sample measured displacement curve, illustrating the generally excellent fit. In these figures, displacements are plotted with respect to the initial head position. The confidence intervals for the rhesus data are greater than those for the humans, reflecting the greater variability in the animal data. This can be seen by comparing Figures 2 and 3. Nonetheless, the displacement equation (2) provides a powerful tool for modelling both human and animal head displacements under conditions of -X impact acceleration.

[4] BMDP Statistical Software, 1985: Nonlinear Regression (P3R), Partial Correlation and Multivariate Regression (P6R), All Possible Subsets Regression (P9R).

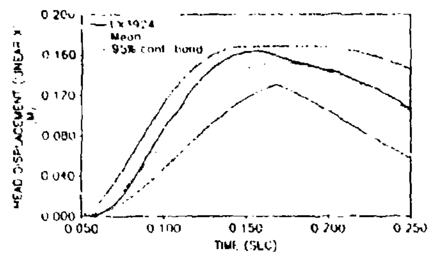

Figure 4: Human X axis linear head displacement

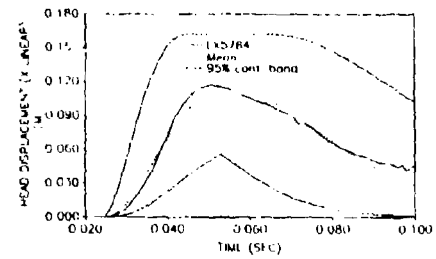

Figure 5: Rhesus X axis linear head displacement

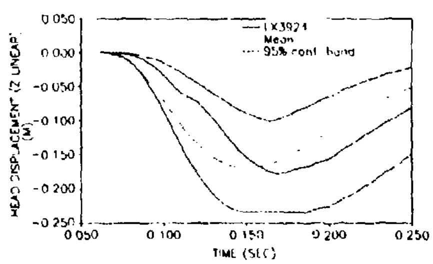

Figure 6: Human Z axis linear head displacement

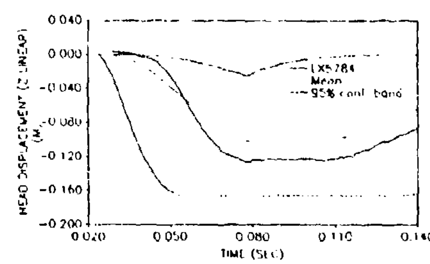

Figure 7: Rhesus Z axis linear head displacement

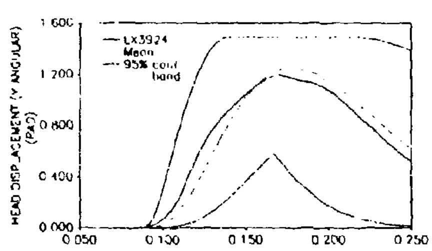

Figure 8: Human Y axis angular head displacement

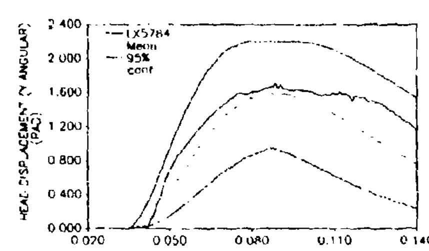

Figure 9: Rhesus Y axis angular head displacement

CONCLUSIONS

The results of this study provide an analytical approach to extrapolating human volunteer kinematics to levels of exposure where injury would be expected. Since the same analytic model (equation (2)) describes rhesus and human head kinematics, the rhesus is an excellent biomechanical surrogate for the human. Previous work indicates a threshold for rhesus head/neck injury at approximately 600 m/s^2 [11,12,14]. To determine the equivalent threshold level for humans, the biomechanical properties of the rhesus head (e.g., mass, center-of-gravity, moments) must be measured. Once these data are obtained, a transform of rhesus dynamics (forces and torques) will provide scaling information enabling injury thresholds to be estimated for humans.

Another important application of these results is to analytically validate anthropomorphic manikins and biomechanical computer models. The model equations can be used to check the displacement equations obtained from these other models over a wide range of g-levels. Similarly, these same techniques can be used to analyze kinematic data obtained from helmeted human volunteers. This analysis could help establish tolerance limits for inertial loading due to the added head mass of helmets and helmet-mounted systems. Efforts continue at NAVBIODYNLAB to extend the validity of this modelling approach to other acceleration directions and to directly address the problem of human injury tolerance under a variety of head-loading conditions.

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APPENDIX I

The matrix solution for parameters to predict linear and angular head displacement is of the form:

$$[q_3, q_2, q_1] = [DAX_1, DAZ_1, PHB_1, PHB^2, PSA, ESV, 1]\ A_{7,3} \tag{3}$$

where $A_{7,3}$ is a 7x3 coefficient matrix with one of the following element structures:

DAX - Human

+1.1784	-0.0881	-0.4068
-0.1832	+0.0421	+0.3450
-1.4179	-0.0223	-0.0363
-11.4412	-0.1249	-0.1279
+0.0219	-0.0002	-0.0004
-0.4200	-0.0044	+0.0051
+1.3265	+0.0435	-0.9147

DAZ - Human

+2.3642	-0.0276	+0.2631
+5.1582	+0.0390	-0.4066
-1.4550	+0.0090	-0.0267
-9.5676	+0.0478	+0.0019
-0.0077	+0.0002	+0.0000
+0.0798	-0.0093	-0.0072
-1.0687	+0.1333	+0.8987

PHB - Human

+6.5058	-0.1277	-2.0798
+15.7618	+0.0795	+1.7881
-8.5138	-0.0832	-0.5042
+30.2184	-0.2680	-0.1416
-0.0509	+0.0000	-0.0024
+0.5487	-0.0086	+0.0886
-14.4521	-0.0592	-1.5279

DAX - Rhesus

-0.4304	-0.0065	-0.6997
-7.4771	+0.0623	-0.4357
+0.0097	-0.0013	-0.0182
+1.2212	-0.0053	-0.0608
+0.0067	+0.0000	-0.0000
-0.2528	-0.0038	+0.0043
+1.5862	+0.0794	-0.6929

DAZ - Rhesus

-14.1133	-0.1105	+0.2585
-17.9610	-0.3769	-1.0371
-1.6668	+0.0546	+0.0150
-1.6750	+0.0539	+0.0020
+0.0003	-0.0001	+0.0000
-0.2560	+0.0058	-0.0079
-4.0467	-0.0213	+0.6862

PHB - Rhesus

-9.1206	+0.0932	+0.1455
+6.4750	-0.7888	+4.4603
+1.5189	+0.0424	-1.4089
+2.0188	+0.0787	+0.3927
+0.0025	+0.0001	-0.0014
-0.2000	-0.0067	+0.1167
-7.8397	+0.2920	-1.8726

ANALYSIS OF THE BIOMECHANIC AND ERGONOMIC ASPECTS OF THE CERVICAL SPINE UNDER LOAD

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SUMMARY

In high performance aircraft complicated loading situations arise, e.g. when the head of the flier is turned backwards and rapid accelerations appear. To obtain more insight in the forces on the cervical spine a spatial biomechanical computer model has been introduced. The research started with the development of a kinematic model which imposes the axes of rotation and mutual position of head and vertebrae in relation to flexion, extension, lateroflexion and torsion.
Subsequently lines of action of muscle forces were introduced as well as external loads acting on the centre of mass of head and helmet born by gravity and by accelerations in different directions. Measurements were carried out of accelerations and head positions during several flights, a.o. representing air combat. Next, with the help of the biomechanical model, forces in vertebrae and muscles could be estimated.
Although in the present stage of the research results of calculations must be interpreted carefully, conclusions can be drawn with respect to sitting posture, head position and helmet devices.
Maximal forces calculated appear to be rather high. However, too few data on failure behaviour exist to draw conclusions as to (long term) detrimental effects.

1. INTRODUCTION

The vast majority of literature on biomechanics deals with bigger joints like hip and knee. With regard to the spine most research is devoted to the lumbar area and to the thoracolumbar spine with respect to scoliosis. Comparatively, research on the cervical spine is very limited, giving a lack of kinematic and anthropometric data. The present study must be weighed in this light.
De Graaf and Ingels[1] and Aghina[2] studied cervical complaints with F-16 pilots and concluded that the origine of discomfort and fatigue closely relates to the amount and duration of the "vertical" acceleration (A_z-acceleration). Also the weight of the helmet, the head position and the fatigue of the aviator play an important role. To these aspects in aerospace medical literature few attention is paid. Most of the studies deal with the origine of acute trauma by unexpected movements by the aircraft and the use of the ejection seat. Also experimental research has been done on maximal sustainable forces in the neck.
The aim of this study is restricted to the analysis of the load on neck structures under high G-load. A biomechanical model is introduced giving access to the calculation of forces in a number of neck muscles and in the joints of the cervical spine. The distribution of forces in a specific joint is not analysed.
Calculations are based on measurements on common flight operations. Furthermore by computer simulation the influence of the helmet and of the positioning of helmet mounted devices on the load in the cervical spine is estimated.

2. BIOMECHANICAL MODEL

2.1 Kinematic model

Head and neck form a kinematic chain consisting of eight links. Every link has six degrees of freedom. The connecting joints restrict the degrees of freedom and the amount of motion. The upper cervical spine consists of atlas and axis which form a loose connection which means that for the positioning and stability of the head muscular forces are always needed. The vertebrae C_3-C_7 possess intervertebral discs. In the following the first simplification regards the assumption that the axes of rotation are located in the middle of the respective joints. In the atlanto occipital joint, however, the axis for lateral flexion is not on the same level as the axis for anteflexion and retroflexion. Furthermore, combining the lower cervical spine to one link (C_3-C_7) leads to Fig. 1 where λ is the angle in the atlanto occipital joint (B), μ is the angle between atlas and axis (C), and ν is the angle between axis and C_3 (E). The origin of the coordinate system is on the caudal-dorsal corner of the vertebra C_7. The length of the link OE is variable, depending on the inclination angle.
The point TC is situated at the top of the clivus corresponding with the location of the centre of gravity of the head. The point AK is point of attachment of dorsal neck muscles at the protuberance occipitalis. The angle of the head with respect to movement in the sagittal plane is called β. The configuration of joints in Fig. 1 represents the neutral position of the head which is taken when the person looks in forward direction with the direction of looking under 15° with the horizontal. Calculations are based on the anthropometric data valid for an average adult man (Dul et al.[3]).
The next part of the kinematic model deals with the relationships of the head and the respective vertebrae while bending forward. A distinction is made between the stages knicking in C_{01} during the first 8° of rotation, and buckling and bending of C_{23} and C_{37} between 8° and 45°. In this phase C_{01} shows a relative retroflexion from +8° to -8°.

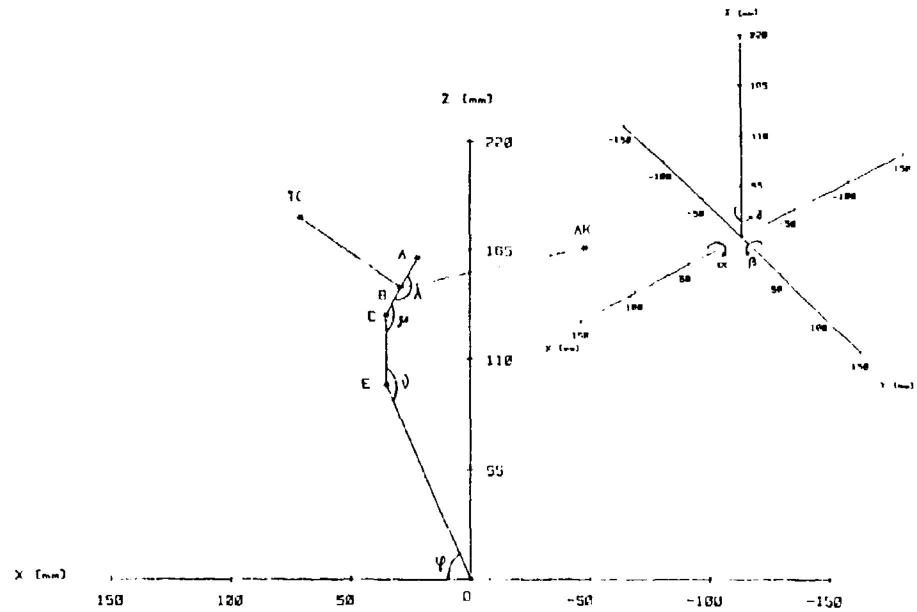


Fig. 1
Kinematic model, neutral position. Ventral side is at the left.
B is axis of rotation in the atlanto occipital joint.
TC is centre of mass of the head and AK point of attachment
of dorsal neck muscles.

So for this stage of antetlexion the following algorithms are assumed in the kinematic model:

$8° < \beta \leq 45°$
$\Delta\lambda = 8° - 16/37\ (\beta - 8°)$ (knicking on C_{01})
$\Delta\mu = 8/37\ (\beta - 8°)$ (buckling on C_{12})
$\Delta\nu = 20/37\ (\beta - 8°)$ (bending on C_{23})
$\Delta\phi = -25/37\ (\beta - 8°)$ (bending on C_{37})
$$OE = OE_1 + \frac{(OE_2 - OE_1)}{37}\ (\beta - 8°)$$

For extension movement, lateral flexion and axial rotation similar algorithms are introduced.

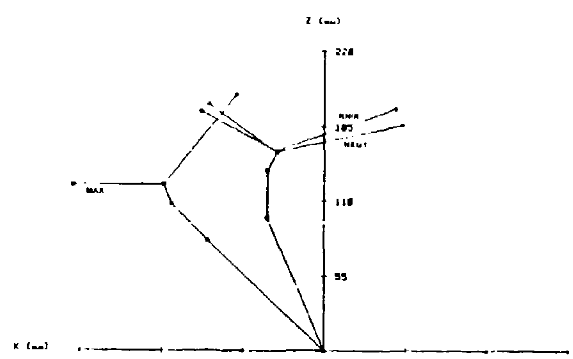


Fig. 2
Forward bending in the model. The first 8° from neutral position, the head knicks in C_{01}. The next phase between 8° and 45° knicking of the head, buckling of the atlas and bending of the lower cervical spine occur.

2.2 Free body diagrams

The next step in biomechanical modelling is the introduction of forces raised by muscles. First the muscles are selected that are supposed to contribute most to the stabilization of head and neck. Next origo and insertion of each muscle had to be estimated, based on anatomy text books and anthropometric literature. In Fig. 3 the free body diagram of the head is given, where FTRR and FTRL are the line of action of respectively the right and the left trapezius muscle. The letters FSCML and FSCMR stand for left and right sternocleidomastoidius muscle. FRC is the m. rectus capitus presented here by one line of action attaching the frontal side of the arc of the atlas. The origo is located on the pars basilaris, point PB.

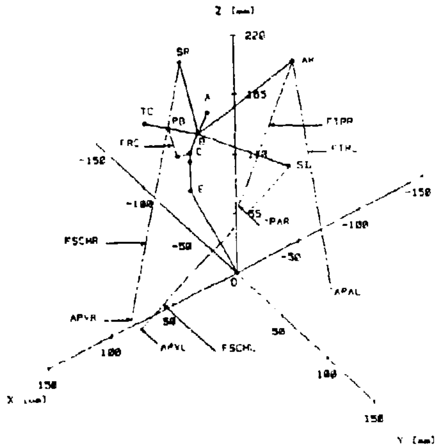


Fig. 3
Free body diagram of the head.
The dotted lines represent lines of action of muscles.

Comparable with the free body diagram of Fig. 3 a separate diagram is made for the atlas and for the lower cervical spine. With regard to the equilibrium of the atlas, special attention is paid to the force in the ligamentum transversum atlantis. As shown in Fig. 4 in forward bend position the force between dens and ligament (FT) can become considerable. In upright position this force is almost absent.
Based on the free body diagrams indicated above, Fig. 5 is the result of the calculation of equilibrium of moments and forces when the head is in neutral position. Here the input parameter is the weight of the head being ca. 45 N. From the calculation of the equilibrium of the head the muscle forces and the total reaction force in the atlanto

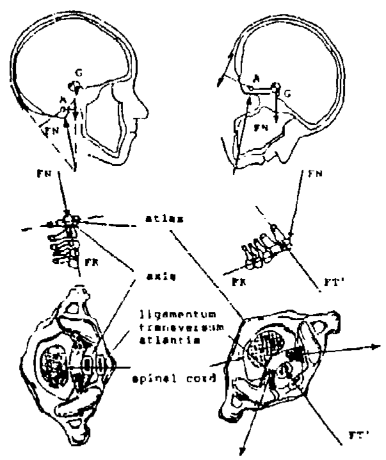


Fig. 4
Force FT in the ligamentum transversum atlantis arising from pressure of the dens, preventing shearing of atlas on axis.

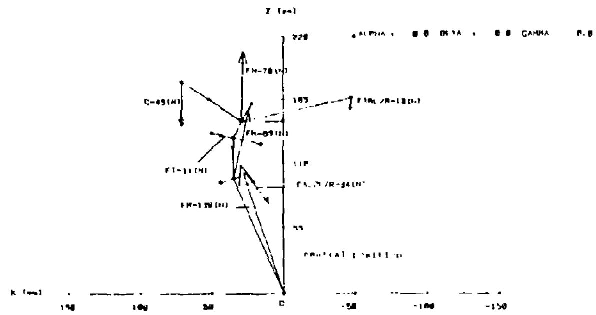


Fig. 5
Forces in muscles and joints when the head is in neutral position.
FN is the joint reaction force in the atlanto occipital joint.
FK in C_{23} and FR in C_7T_1.
The joint reaction forces are in the sagittal plane, the muscle forces are not.

occipital joint are derived. Next the equilibrium of the atlas is calculated with the atlanto occipital joint reaction force calculated earlier as an input parameter. The same procedure followed for the lower cervical spine. Due to the number of muscle forces the static model is over-determined. Therefore a basic optimalisation algorithm is used by which three muscle groups are selected for every link leading to the smallest joint reaction force (bone contact force).
Starting from Fig. 5 calculations can be performed for flexion, extension, lateroflexion, torsion and combinations of these different rotations. As an example in Fig. 6A the joint reaction forces are calculated for the maximal excursion in axial rotation and in Fig. 6B a plot of some muscle forces is given.
A sensitivity analysis has been executed to determine which parameters influence the results most. In the neutral position and the extreme positions all parameters have been varied one after the other up to a deviation of 10%. Those parameters are understood to be critical when the influence in the results became greater than ± 10%. As expected it appeared that geometry in neutral position showed most influence. With the parameter ϕ_0 the joint reaction force FR showed a deviation up to 25%. This force showed a deviation of 60% when to all parameters a deviation of 10% was given. So the sensitivity of the model for geometric data is fairly great, emphasizing the importance of reliable anthropometric data.

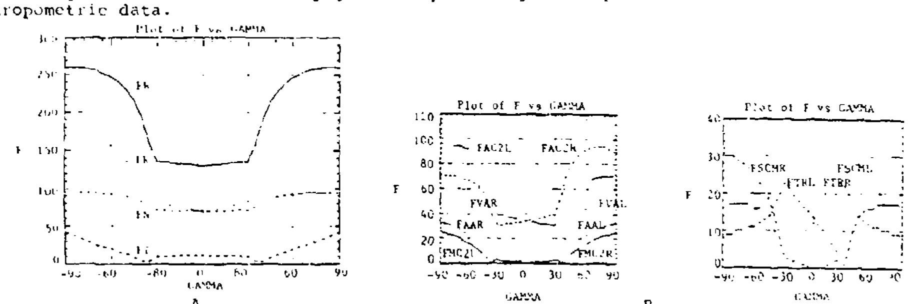


Fig. 6
A: Relation between joint reaction forces and axial rotation.
B: Relation between muscle forces and axial rotation.
F in N, gamma in degrees.

3. METHODS AND MATERIALS

Measurements were executed with one pilot performing a number of normal flight operations with an instrumented F-16. For determination of the position of the head during flight the standard F-16 video camera was turned 180°. On the helmet and the shoulders markers were fixed allowing for three-dimensional reconstruction of position. To determine the direction and the magnitude of the acceleration vector acting on the head three accelerometers were located on the helmet in a perpendicular coordinate system. Data acquisition and calculations were executed by the Dutch National Aerospace Laboratory. The inaccuracy of the head position angles obtained by the method described is estimated on 3-4°. The accuracy of the accelerometers is approximately 0.2 m/s^2.
Four flights were performed with a pilot of average posture. From these flights 330 situations were randomly selected and analysed.
Figure 7 gives the spectrum of the vertical acceleration A_z. It appears that the value of $+G_z$ remains below 2 $+G_z$ during 49% of the total time of the flight. During 5% of the time values of 7-9 $+G_z$ appear.
With respect to the four flights an analysis is made regarding the question whether relations exist between different parameters.

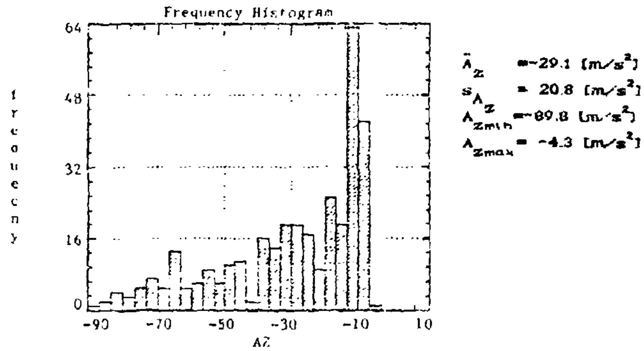


Fig. 7
Frequency distribution of A_z during four flights.
$+G_z$ values below 2 $+G_z$ during 49% of the time.
Above 7 $+G_z$ during 5% of the time.

Summarizing the following can be mentioned:

- When the vertical acceleration (A_z) increases (in a curve) the acceleration in forward-backward direction increases too (braking) approximately according to $A_x = -0.12\ A_z\ (m/s^2)$.
- When A_z increases, axial rotation with great excursion coupled to lateroflexion in the direction of rotation (looking backward in air combat simulation) occurs relatively more often, approximately according to $\alpha = 0.25\ \gamma$. This finding may be a support for the biomechanical model introduced, because this relation approximately corresponds with the minimal forces as calculated with the help of the model (the "valley" in Fig. 8).

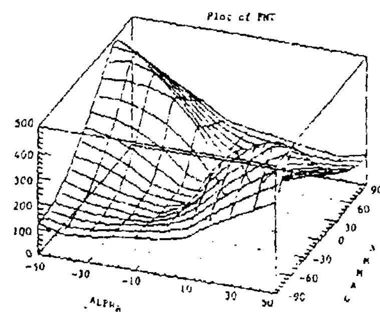


Fig. 8
Forces in the atlanto occipital joint as calculated in relation to values of lateroflexion (α) and axial rotation (γ).
The "valley" indicates the region of minimal forces.

4. RESULTS

In Fig. 9 calculations based on the observed neutral position of the head and neck while flying straight forward or in a moderate curve and looking forward is given. Here the weight of the helmet (GH = 18 N) is added. Comparison with Fig. 3 shows a steeper position of the lower cervical spine and forces in the same order of magnitude.

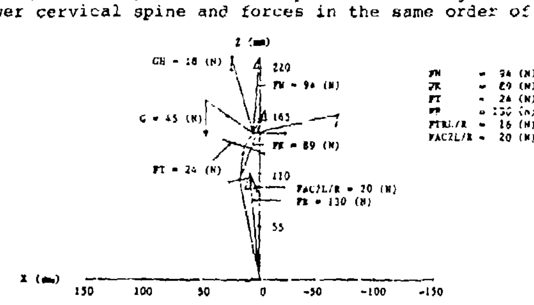
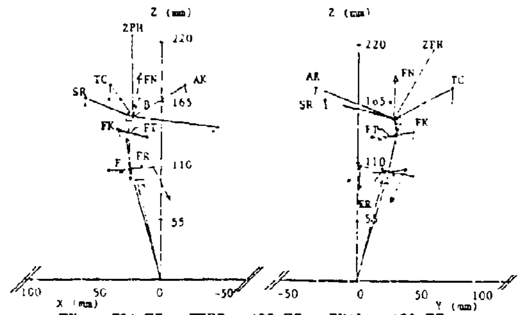


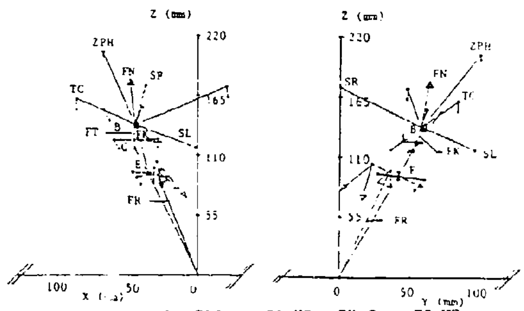
Fig. 9
Forces calculated for the F-16 neutral position.



FN = 734 [N]	FTRR =138 [N]	FVAL =123 [N]
FK = 934 [N]	FSCML =136 [N]	FAC2L =625 [N]
FT = 205 [N]	FSCMR =103 [N]	FAC2R =451 [N]
FR =1988 [N]	FAAL = 48 [N]	FMC2R =316 [N]

Fig. 10
The cervical spine under high $+G_z$-load (6 $+G_z$).
Combination of axial rotation and moderate forward and lateroflexion.

Furthermore the following two situations, which often occur and can be considered quite heavy, are selected. Figure 10 illustrates situations in the interval between 5 and 7 $+G_z$ with reasonable great axial rotations (γ = 63°) with moderate lateroflexion (α = 16°) and forward flexion (β = 18°). Due to the high value of $+G_z$ (6 $+G_z$) the load by head and helmet weight is increased to 377 N and with A_x = 7 m/s^2 the joint reaction forces and muscle forces become considerable. Based on data concerning the forces that can be sustained by muscles during a certain period of time it can be concluded from these figures that this posture can be taken only 10 to 30 seconds approximately without discomfort, i.e. by persons without special training.



FN =187 [N]	FRC = 58 [N]	FVAR = 59 [N]
FK =231 [N]	FTRL = 44 [N]	FAC2L = 67 [N]
FT = 29 [N]	FSCMR = 31 [N]	FAC2R =177 [N]
FR =464 [N]	FAAR = 51 [N]	FMC2R =143 [N]

Fig. 11
Adjusting the aircraft computer in forward flexion.
A long duration makes this posture uncomfortable.

In Fig. 11 a less extreme situation is presented. Pilots, however, experience this posture being uncomfortable because it concerns the adjustment of the aircraft computer in forward flexion during a longer period of time (up to 15 minutes). The calculations indicate that, for instance, the force in the left trapezius muscle (FTRL = 44 N in Fig. 11) can only be sustained approximately 10 minutes without fatigue.
In Table I, a comparison of the load situations described is made with respect to the neutral position by dividing the loads calculated by the values corresponding with the neutral position without helmet (situation 0). Case 3 is added with a position similar to case 2. However, the forces are greater because of A_x = 10 m/s^2 and A_z = 8 $+G_z$. Separately a number of loading situations has also been calculated without the weight of the helmet. In average, it appeared that addition of the weight of a helmet increases the joint reaction forces with a factor 1.3. This contribution even increases with the addition of helmet mounted devices. To investigate the influence of the latter, a mass of 0.58 kg was introduced in the model. Furthermore the effect of different locations of this extra mass on the forces in the neck was calculated.
In Fig. 12 the results are summarized. A_z is $+G_z$ and A_x is 1 m/s^2. In the diagrams for lateroflexion (Fig. 12A), flexion and extension (Fig. 12B) and axial rotation (Fig. 12C) index 1 indicates the F-16 flight situation. With index 2 the loads increase by addition of the helmet mounted device. The lines with index 3 are found when an additional

situation	"head" load	FN	FK	FR	FT
0	45	70	68	130	11 ← N
1	1.40	1.34	1.31	1.00	2.18
2	8.47	10.49	13.74	15.29	18.64
3	11.29	14.16	19.10	20.95	27.09
4	1.40	2.67	3.40	3.72	2.64

Table I
Joint reaction forces divided by those calculated in the neutral position without helmet (0).
1 = neutral position in F-16; 2 = Fig. 10;
3 = position comparable with Fig. 10; 4 = Fig. 11.

counterweight of 0.44 kg mass is added at the dorsal side of the helmet, as tested by a pilot. Finally, index 4 is found by computer simulation of a helmet of only 0.8 kg mass and a centre of mass as far behind axis B as it is in front of this axis in Fig. 9. From the results it may be concluded that notwithstanding an increased mass of 0.44 kg the addition of this counterweight in an appropriate place positively influences the load on the cervical spine, especially in axial rotation. However, in reality every addition of mass should be avoided if possible.

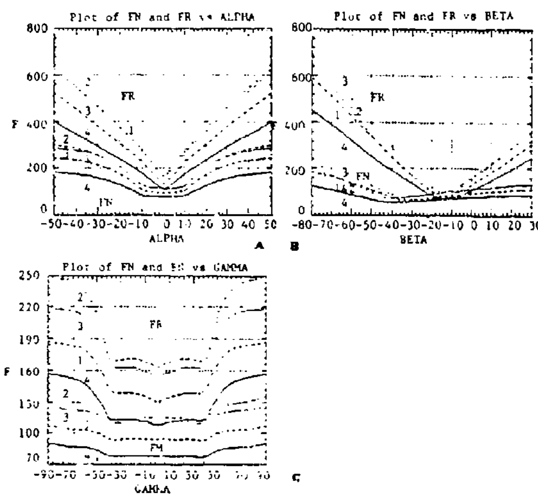


Fig. 12
Effect of helmet and helmet mounted devices on atlanto occipital joint reaction force FN and joint force C_7T_1.
1: neutral F-16. 2: addition of helmet mounted device.
3: addition of counterweight. 4: light helmet (0.8 kg) with backward location of centre of gravity

5. CONCLUSIONS

When drawing conclusions it must be emphasized that the model is a simplification of reality and that reliable anthropometric data hardly exist. Although it is difficult to verify the model with respect to demonstration of real existing forces, we could obtain indications that the order of magnitude is correct. So the approach followed leads to results, be it inaccurate. The most reliable conclusions may be drawn with respect to the comparison of different situations. So emphasizing on the reserved use of the results of this study it may be concluded that:

a) With increasing vertical acceleration (A_z) the forward-backward acceleration (A_x) also increases, indicating that the aircraft decelerates.
b) With low values of $+G_z$ a neutral position occurs or slight extension while with increasing $+G_z$, this turns to slight flexion. With high $+G_z$ accelerations the pilot seldom looks upward while relatively often great axial rotations occur when looking backward over the shoulders.
c) A relation is found between the lateroflexion of the head (α) and its axial rotation (γ) according to: $\alpha = 0.25\ \gamma$. This relation measured during flight corresponds with minimum values as calculated for joint reaction forces in the biomechanical model.
d) The mass of the helmet is big in proportion to the mass of the head leading to forces in the neck being 1.3 up to 1.5 greater.
e) The position in the seat of the F-16 seems to be favourable because it decreases the lordosis of the cervical spine and as such the forces in the lower neck.
f) The combination of posture and high G-load can multiply the load in the atlanto occipital joint 14 times. In the lower cervical area this multiplying factor can be 21,

occurring when under high G-load the head is rotated extremely.

g) Addition of an extra counterweight on the helmet to balance the influence of a helmet mounted device can decrease the load on the cervical spine, even if the total mass incr ss by doing so. Improvement of the position of the mass centre of gravity of heln. and helmet mounted devices can lower the load on the spine.

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EFFECTS OF HEAD MOUNTED DEVICES ON HEAD-NECK DYNAMIC RESPONSE TO $+G_Z$ ACCELERATIONS

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SUMMARY

An investigation is described which addresses the inertial loading effects of Head Mounted Devices (HMD) on aviator head-neck-spine dynamic response during high $+G_Z$ acceleration exposure. The primary objectives of this study were to develop a methodology which could be used to establish limits on HMD inertial properties and to apply this methodology to the evaluation of the severity of the internal loads -- occurring in the neck and upper spine -- associated with certain specific HMD ensembles. This paper describes how the Head-Spine Model (HSM), a highly discretized, 3-D mathematical representation of the human head-spine-torso structure, was used to: 1) establish a set of baseline response criteria (BRC); 2) establish a preliminary methodology for setting limits on HMD inertial properties; and 3) evaluate the severity of the loading associated with possible chemical defense (CD) ensembles.

INTRODUCTION

The investigation described in this paper was part of a more encompassing program which is being conducted at the Harry G. Armstrong Aerospace Medical Research Laboratory (AAMRL) located at Wright-Patterson Air Force Base, Ohio. This program has as its overall goals the development of design guidelines for limiting the inertial properties of HMD for various dynamic environments and the establishment and implementation of methodologies that will provide accurate measurements of the inertial properties and evaluations of the inertial loading severities associated with existing or planned HMD. Motivation for this program stems from the increasing emphasis on the use of the aviator's head and/or helmet as platforms for protective and/or performance enhancement equipment such as chemical defense gear or night vision enhancement systems.

While such equipment indeed increases crewmember protection and enhances performance, organizations within the United States Air Force, Navy and Army are nonetheless concerned about the potentially adverse effects associated with HMD (1). These adverse effects arise from the c.g. (center of gravity) shifts, usually anteriorly, and increased loading, on the neck and upper spine, produced by HMD. They include excessive helmet motion relative to the head, neck muscle fatigue and, in high G environments, a potentially significant increase in the likelihood of severe injury to the neck and upper spine. Designers of HMD are endeavoring to minimize these systems' weights and c.g. distances from the head c.g. (see e.g., (2)). They are having to do so, however, without the aid of well established quantitative guidelines based on, e.g., neck and upper spine load limitations.

AAMRL's Program, which seeks to establish such quantitative guidelines, has involved both analytical and experimental aspects. The experimental work has considered the measurement of the inertial properties -- mass, inertia tensor and c.g. location -- of specific HMD, using an automated "mass properties measurement system", and the conducting of a series of $+G_Z$ impact tests on AAMRL's six inch "HYGE" vertical impact facility. The impact tests focussed on a Hybrid III manikin head-neck structure plus five specific helmet plus mask combinations, four of which represent possible CD configurations. The analytical investigation, which is emphasized in this paper, used the Head-Spine Model (HSM), a highly discretized, 3-D mathematical representation of the human head-spine-torso structure, to: 1) establish a set of baseline response criteria (BRC); 2) establish preliminary guidelines for limiting HMD inertial properties; and 3) evaluate the severity of the inertial loading associated with the five helmet plus mask configurations.

The experimental portions of AAMRL's program, along with the analytical investigation, are discussed in detail in AAMRL-TR-88-044 (3). Some aspects of this program have also been described in references (4) and (5).

MODEL DESCRIPTION

The HSM is a three-dimensional mathematical model describing the mechanical behavior, in terms of system kinematics and internal loads, of the human head-spine-torso structure. Its fully three-dimensional formulation is just one of the features which significantly distinguishes it from earlier such models. The HSM consists of two distinct components: a general purpose computer program for the dynamic analysis of three-dimensional structures; and a data base containing inertial, material, geometric and connectivity data describing the head-spine-torso structure as well as other information descriptive of the specific problem and output to be generated. The HSM has been described previously by Belytschko, et

al. (6), Belytschko and Privitzer (7), Privitzer and Belytschko (8), and Privitzer (9), thus, only a very brief description will be given here.

Figure 1 depicts mid-sagittal (X-Z) and frontal (Y-Z) plane views and also an oblique view of the initial HSM geometry. These computer graphics generated plots show only those components of the model whose local geometries are treated as constant: the head, pelvis, the vertebrae of the cervical and thoracolumbar (TL) spines and the elements of the rib cage. None of the deformable elements representing connective tissues are shown. This is actually the most complex (in terms of the number of degrees of freedom) version of the HSM and, in the interest of computational efficiency, is rarely used for studies involving large numbers of simulations.

The version of the HSM used for the study reported herein, models the neck with two parallel 3-D beam elements. One of these beam elements has nonlinear viscoelastic axial load-deformation behavior and linear viscoelastic bending behavior and is used to represent the cervical spine. The other neck beam element has only nonlinear bending behavior, i.e., it provides no resistance to purely axial deformations, and is used to account for the nonlinear stiffening effects of the soft tissue under large neck bending deformations. This element is also used to account for chin-chest contact under large neck bending deformations. The secondary loading path and nonlinear stiffening effects of the viscera-abdominal wall-diaphragm-rib cage system are accounted for with a column of nonlinear bending elements which roughly parallels the spinal column. These elements interconnect the c.g.'s of the torso segments and develop significant bending resistance only in the case of large relative rotations between adjacent segments.

The HSM's geometry is defined by the global coordinates of points identified as primary and secondary nodes and by triads of unit vectors giving the orientations of the rigid bodies. The primary nodes correspond to the c.g.s of rigid bodies and also serve as the origins of the local coordinates attached to the rigid bodies and coinciding with their principal axes of inertia. Inertial properties are specified in terms of each body's mass and principal mass moments of inertia. The secondary nodes define some local geometric features, such as vertebral geometries, and serve primarily as attachment points for the deformable elements representing the various connective tissues. The deformable elements of the HSM version employed in this study include beam elements used, e.g., to model the intervertebral discs and spring elements used, e.g. to model the spinal ligaments. Deformable element equilibrium equations are given by:

axial forces --

$$f_{xJ} = k_x \left(\delta + \frac{2\mu_a}{\beta_a}\dot{\delta}\right), \quad f_{xI} = -f_{xJ}; \tag{1}$$

torsional moments --

$$M_{xJ} = \frac{GJ}{L}\,\theta_{xJI}, \quad M_{xI} = -M_{xJ}; \tag{2}$$

bending moments --

$$\begin{Bmatrix} M_{qI} \\ M_{qJ} \end{Bmatrix} = \frac{k_q}{1+\phi_q}\begin{bmatrix} 4+\phi_q & 2-\phi_q \\ 2-\phi_q & 4+\phi_q \end{bmatrix}\left(\begin{Bmatrix} \theta_{qI} \\ \theta_{qJ} \end{Bmatrix} + \frac{2\mu_b}{\beta_b}\begin{Bmatrix} \theta_{qI} \\ \theta_{qJ} \end{Bmatrix}\right) \tag{3}$$

and shear loads --

$$\begin{aligned} f_{yI} &= \frac{M_{zI} + M_{zJ}}{L}, \quad f_{yJ} = -f_{yI} \\ f_{zI} &= -\frac{M_{yI} + M_{yJ}}{L}, \quad f_{zJ} = -f_{zI}. \end{aligned} \tag{4}$$

All quantities in equations (1) through (4) are defined with respect to local element coordinate systems which are referred to as rigid-convected systems since they are attached to the elements and move with them through space. I and J refer to the endpoints or nodes of an element. In equation (1);

- x is directed along the length of the element from node I to J,
- k_x = axial stiffness (can be nonlinear),
- δ = deformation,
- $\dot{\delta}$ = deformation rate,
- μ_a = fraction critical damping,
- β_a = global axial circular frequency to be damped.

In equation (2);

G = shear modulus,
J = polar moment of inertia of the cross-sectional area,
L = element length,
$\theta_{x_{JI}}$ = $\theta_{x_J} - \theta_{x_I}$ = torsional deformation.

In equation (3);

q refers to either the y or z axes,
k_q = bending stiffness (can be nonlinear),
θ_{qI}, θ_{qJ} = bending deformations,
$\dot{\theta}_{qI}, \dot{\theta}_{qJ}$ = bending deformation rates,
μ_b = fraction critical damping,
β_b = global bending circular frequency to be damped,
ϕ_q = shear deformation parameter,

$$= \frac{12\,E\,I_q}{G\,A_q\,L^2}$$

E = modulus of elasticity,
I_q = second moment of the cross-sectional area about q,
A_q = area effective in shear.

Material nonlinearities are incorporated by defining k_x and k_q to be nonlinear functions of deformations.

In addition to the deformable elements representing the internal connective tissues, a system of spring elements is used to model a restraint system and viscoelastic surfaces are used to represent interaction surfaces such as an ejection seatback. The experimental and analytical bases for the selection of the HSM geometry and inertial and material properties are described in detail in references (6), (7), and (10) through (13).

The HSM computer program uses an explicit scheme for the numerical time integration of the nonlinear equations of motion for model kinematics. the approach used requires no matrix inversions. All element quantities are computed at the element level, i.e., with respect to the rigid-convected coordinates, $\hat{x}_k$. After the element by element computations have determined the element nodal loads, they are transformed and assembled into a global internal force array, F^{int} (defined in the global coordinates, X_k) and into internal moment arays, $\bar{M}_I^{int}$ (the components of which are defined with respect to the various body systems, $\bar{x}_k$), corresponding to each primary node (rigid body), I. The components of F^{int} are then used in the computations for translational kinematics via Newton's Second Law while the components of the $\bar{M}_I^{int}$ are used in the computations for rotational kinematics via Euler's Equations of Motion for each rigid body. The procedure is described in detail by Belytschko, et al.,.(14)

Spinal Injury Function and Neck Injury Parameter

The HSM has a spinal injury prediction capability, referred to as the Spinal Injury Function (SIF), which addresses the predominant ejection acceleration as well as general vertical impact acceleration induced spinal injury mode; vertebral body compressive failure resulting from combined axial compression and bending loads. It is given by:

$$SIF_V = \left\{ \left| \frac{P}{P^*} \right| + \max\left[\left| \frac{M_x}{M_x^*} \right| , \left| \frac{M_y}{M_y^*} \right| \right] \right\}_V^{max} \qquad (5)$$

where V = vertebral level of the thoracolumbar (TL) spine; P, M_x and M_y are simulation computed instantaneous equilibrium values of the compressive load and the local lateral and AP bending moments, respectively; and P^*, M_x^* and M_y^* are the corresponding failure levels. The P^* are based on rate dependent axial compression load-deformation data (to failure) (15) and (16). The corresponding data for the M_x^* and M_y^* were found to be insufficient. These were thus derived from the P^* through the use of relationships based on assumptions on vertebral body geometry and material distribution (3). The SIF, as given by equation (5), represents the ratio of extreme fiber compressive stress to a failure or limiting value. Thus, assuming that the compressive limiting stresses are normally distributed, a value of SIF = 1 at any vertebral level V of the TL spine is taken to correspond to a 50% likelihood of vertebral body compressive failure due to combined axial compression and bending at that level.

A Neck Injury Parameter (NIP) was developed, as part of this investigation, to provide an injury prediction feature for the neck similar to the SIF for the TL spine. The NIP is given by:

$$f_N = \left\{ \left| \frac{P}{P^*} \right| + \max\left[\left| \frac{Mx^{avg}}{M^*} \right| , \left| \frac{My^{avg}}{M^*} \right| \right] \right\}^{max} \qquad (6)$$

where

$$Mx^{avg} = \frac{Mx^i + Mx^j}{2} \; ; \quad My^{avg} = \frac{My^i + My^j}{2} \; ;$$

M_x and M_y refer to local lateral and A-P bending moments respectively, and the supercripts i and j refer to nodes i and j of the neck beam element and correspond to the C7-T1 and Head-C1 junctures, respectively. As is the case for the SIF, the P, M_x^{avg} and M_y^{avg} in equation (6) refer to simulation computed instantaneous equilibrium values of the compressive load and the local lateral and A-P bending moments while the P* and M* are the corresponding failure levels. Because of the approximately elliptical cross-sectional geometry of the vertebral bodies, the lateral and A-P limit bending moments for the SIF are not equal -- the lateral limit bending moments are generally larger than the A-P limit bending moments since the lateral vertebral body diameter is typically larger than the A-P diameter. For the NIP, however, it was assumed that the lateral and A-P limit bending moments are equal. Figure 2 shows the limit loads for the SIF and f_N plotted versus vertebral level (L5 through T1) for the SIF and a single point (corresponding roughly to the middle of the cervical spine) for the neck. Note that the limit loads for the neck were extrapolated from those for the TL spine.

Validation of the HSM has been pursued at AAMRL for a number of years (17). It has involved comparisons of model predictions with data obtained from experimental programs and also spinal compressive injury statistics compiled from operational ejection data. HSM dynamic response predictions have been found to compare well with data obtained from experiments with human volunteers ((7), (8), and (18)). Comparisons of HSM-SIF predictions with operational ejection injury statistics appear to be reasonable with respect to both predicted injurious acceleration profiles and spinal injury locations. Note again that the vertebral body axial compression failure levels used by the SIF, i.e., the P* in Equation (5) are based directly on data obtained from rate dependent axial compression load-deformation experiments with human vertebral bodies.

APPROACH

Our approach to the analytical study began with the use of the HSM to establish a set of limiting or baseline response criteria (BRC). These were HSM neck and spinal response predictions from a simulation with a moderate risk + G_z half-sine acceleration exposure. Following this, HSM ejection simulations were run for different configurations of generic encumbering devices (point masses). Guidelines for setting limits on encumbrance mass and location were then established by comparing HSM neck and spinal response predictions from these simulations to the BRC. Finally, a series of HSM ejection simulations was run for the specific helmet and mask combinations considered in the experimental part of the program. The performances of these ensembles were evaluated against the HSM established guidelines.

Baseline Response Criteria

The response parameters of primary concern in this study were the NIP for the neck and the SIF for the TL spine. Thus, in order to quantify the inertial loading effects of HMD, we required a set of limiting or baseline response criteria (BRC) for these parameters. Ideally, such criteria should be based directly on appropriate experimentally measured data. For the lower TL spine, some such data do indeed exist, e.g., those on which the P* in equation (5) are based. As already mentioned, however, similar such data for the TL spine limit bending moments, M_x* and M_y* were insufficient. This was also true for any such data for the cervical spine. Note that what we desired for the cervical spine were limiting compression loads and bending moments at specific locations, such as specific cervical vertebrae, not limiting loads deduced from experiments with human volunteers or cadavers.

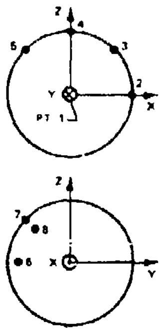
Because of this lack of appropriate experimental data, it was decided to base the BRC on the HSM's response to a moderately severe whole body + G_z acceleration exposure. The specific profile is a 17G peak, 300 ms duration half-sine prescribed to act at the HSM pelvis c.g. and the seatback. This moderate risk exposure is based on the whole body acceleration tolerance criteria established by AAMRL for the Aerospace Medical Division's CREST (Crew Escape Systems Technologies) Program (19). The term moderate risk implies a 5% probability of spinal injury. Figure 3 shows the NIP and SIF as well as the ratios P/P* and M/M* from the HSM baseline simulation, i.e. the HSM predicted response (in terms of spinal loads) to the moderate risk +G_z half-sine exposure. Note that only one bending moment ratio is plotted for the TL spine since the response for this simulation was symmetric about the mid-sagittal (X-Z) plane. Thus the M/M* for the TL spine refer to A-P bending. The f_N and SIF given in Figure 3 are the BRC.

Ejection Simulations with Generic Encumbrances

Following the establishment of the BRC, an extensive matrix of ejection simulations was run in which generic encumbrances, i.e., point masses of 1, 2 and 3 kg, were located at 8 different points on the surface of the helmet (see Table 1). The simulations plus the nomenclature used to identify them are listed in Table 2. Note that while Tables 1 and 2

include both symmetric and asymmetric configurations, only the symmetric cases are discussed in this paper. All of the simulations, including the 17G, 300 ms half-sine exposure included the effects of a generic helmet having a mass of 1 kg, principal mass moments of inertia of 100 kg-cm^2 and with its c.g. assumed to be coincident with that of the head. The helmet was also assumed to move with the head thus its inertial properties were added directly to those of the head. Similarly, the point masses were also assumed to move with the head/helmet, thus the inertial properties of a head/helmet/point mass system were calculated with respect to the shifted c.g. of the entire system.

Table 1
COORDINATES OF POINT MASS LOCATIONS IN HEAD/HELMET LOCAL SYSTEM



POINT MASS LOCATION	COORDINATES IN HEAD HELMET LOCAL SYSTEM (cm) X	Y	Z
1	0	0	0
2	12	0	0
3	$12/\sqrt{2}$	0	$12/\sqrt{2}$
4	0	0	12
5	$-12/\sqrt{2}$	0	$12/\sqrt{2}$
6	$12/\sqrt{2}$	$-12/\sqrt{2}$	0
7	0	$-12/\sqrt{2}$	$12/\sqrt{2}$
8	$12/\sqrt{3}$	$-12/\sqrt{3}$	$12/\sqrt{3}$

Table 2
NOMENCLATURE FOR EJECTION SIMULATIONS WITH GENERIC ENCUMBRANCES

ID	POINT MASS LOC	DEFINITION
BGH	–	BASELINE, GENERIC HELMET
AGH	–	ACES II, GENERIC HELMET
CG1, 2 & 3	1	1, 2 & 3 kg @ HEAD/HELMET (H/H) C.G.
A1, 2 & 3	2	1, 2 & 3 kg @ H/H ANTERIOR PT.
AS1, 2 & 3	3	1, 2 & 3 kg @ H/H ANTERIOR-SUPERIOR PT.
S1, 2 & 3	4	1, 2 & 3 kg @ H/H SUPERIOR PT
ASPS1, 2, 3 & 4	3 & 5	0.5, 1, 1.5 & 2 kg @ H/H ANTERIOR SUPERIOR + POSTERIOR-SUPERIOR PTS.
AR1, 2 & 3	6	1, 2 & 3 kg @ H/H ANTERIOR-RIGHT PT.
RS1, 2 & 3	7	1, 2 & 3 kg @ H/H RIGHT SUPERIOR PT
ARS1, 2 & 3	8	1, 2 & 3 kg @ H/H ANTERIOR-RIGHT SUPERIOR PT

The ejection acceleration exposure chosen for these simulations was a nominal ACES II catapult plus rocket acceleration profile with a 12 G peak and a time to peak of 140 ms (20). The 17 G, 300 ms half-sine and the ACES II acceleration profiles are plotted in Figure 4. The HSM head-neck ranges of motion are similar for both exposures. In fact, the primary criteria for the selection of the baseline exposure were 1) that it be moderate risk, 2) that it be representative of experimentally attainable exposures and 3) that it produce a head-neck range of motion similar to that associated with the nominal ACES II profile.

Figure 5 compares the HSM predicted head - neck - TL spine kinematic responses from simulations BGH, AGH and AS3 (see Table 2 for simulation nomenclature). Shown are mid-sagittal (X-Z) plane configurations at 150, 200 and 250 ms. These configurations are representative of the range of kinematic responses associated with all of the symmetric simulations. Only those components of the model whose local geometries remain constant are plotted by the HSM's plotting software. Thus, in this case, the head or the head/helmet/encumbrance system, the pelvis and the vertebrae of the TL spine are plotted while the deformable elements of the TL spine and the neck beams are not. A reasonable estimate of the deformed geometry of the beam element representing the cervical spine can, however, be obtained from the kinematics of the head (or head/helmet/encumbrance system) and T1 -- hence the dashed curve approximating the deformed geometry of this element in the 200 ms configurations. Kinematically speaking (and also qualitatively), it is quite apparent that the AGH response is less severe than the BGH response, while the AS3 response is more severe. Figure 6 compares head mid-sagittal plane rotations from simulations BGH, AGH and AS3 while Figure 7 compares T1 rotation time histories. The BGH and AGH responses are quite similar except for the higher magnitude of the BGH head and T1 rotations resulting from the higher peak acceleration of the 17G, 300 ms half-sine exposure.

Ejection Simulations with Specific HMD

Following the completion of the HSM ejection simulations with the generic HMD, additional ejection simulations were run which incorporated five specific helmet plus mask combinations used in the experimental portion of this program. Two helmets were considered; a "pilot's" helmet (HGU-55/P) and a "flyer's" helmet (HGU-39/P). Three masks were considered: a pilot/crewmember oxygen mask (MBU-12/P) and two chemical-biological-oxygen (CBO) masks (MBU-13/P and AR-5). The inertial properties of the helmet plus mask configurations were obtained in the experimental portion of the program.

Table 3 lists the five specific helmet plus mask combinations and the inertial properties of the complete helmet + mask + head systems. These data are also included for the generic encumbrance configurations for simul tions AGH, CG1, CG2 and CG3 for comparison purposes. Since the simulations were symmetric with respect to the mid-sagittal (X-Z) plane -- as were the inertial properties (at least nearly so) of the specific HMD -- the relevant inertial properties are mass, principal mass moment of inertia about the lateral axis through the system c.g. (I_y), and the X and Z locations of the system c.g. with respect to the unencumbered head c.g. The HSM unencumbered head has a mass of 4.38 kg and an I_y of 233.0 kg-cm^2 compared to 4.54 kg and 240.0 kg-cm^2, respectively, for the Hybrid III head. The parameters R_N and R_S(T1) will be described in the next section.

Table 3
SPECIFIC HMD CONFIGURATIONS, INERTIAL PROPERTIES, AND R_N (NIP RATIOS) AND R_S (T1) (SIF(T1) RATIOS) FROM SIMULATIONS WITH SPECIFIC HMD

CONFIGUR-ATION	HELMET	MASK OR POINT MASS	SYSTEM[1] MASS (kg)	SYSTEM[1] I_y (kg-cm²)	SYSTEM C.G. LOCATION (cm)[1] x	z	R_N	R_S (T1)
1	HGU-55/P	MBU-12/P[2]	5.75	383.	-0.43	-0.34	0.91	0.61
2	HGU-55/P	MBU-13/P[3]	6.11	424.	-0.16	-0.27	0.90	0.67
3	HGU-55/P	AR-5[3]	6.19	486.	-0.60	-1.18	0.90	0.65
4	HGU-39/P	MBU-13/P	6.60	513.	-0.38	-0.63	0.95	0.70
5	HGU-39/P	AR-5	6.68	499.	-0.76	-0.94	0.99	0.66
AGH	"Generic"	—	5.38	333.	0.0	0.0	0.84	0.59
CG1	"Generic"	1.0	6.38	333.	0.0	0.0	0.93	0.71
CG2	"Generic"	2.0	7.38	333.	0.0	0.0	1.0	0.82
CG3	"Generic"	3.0	8.38	333.	0.0	0.0	1.02	0.94

NOTES
(1) helmet + mask + head
(2) oxygen mask
(3) chemical-biological-oxygen (CBO) mask

RESULTS

The inertial loading effects of first the generic encumbering devices and then the specific HMD were evaluated by comparing the HSM NIP and SIF predictions, from the ejection simulations with those devices, to the BRC, i.e., the NIP and SIF predictions from the simulation with the 17G, 300 ms half-sine exposure (simulation BGH - Baseline with Generic Helmet). These comparisons were accomplished by dividing the NIP and SIF from the ejection simulations by the corresponding BRC values and then plotting these ratios versus spinal level. Thus when any of these ratios exceed 1.0, the corresponding BRC or limiting value is exceeded.

Results from Ejection Simulations with Generic HMD

Figures 8 and 9 show the effects of varying a point mass from 0 to 3 kg at locations 2 and 3, the head/generic helmet anterior and anterior - superior points, respectively. Figure 10 shows the NIP and SIF ratios as functions of the location of a 2 kg point mass - actually case 5 corresponds to a 1 kg point mass located at both sites 3 and 5. Results are plotted for the neck and vertebral levels T1 through T6. The lower levels (T7 through L5) are not included because the inertial loading effects of the point masses were found to decrease with increasingly lower vertebral level. It was also found that, for all cases of interest, the largest ratios involved the NIP and SIF(T1) (SIF at T1). This observation indicates that we actually do not need to consider 18 BRC (the 17 SIF plus the NIP). Rather, we can focus on two parameters in particular: the SIF ratio at T1, which for convenience will be referred to as $R_S(T1)$; and the NIP ratio, which will be referred to as R_N, i.e.,

$$R_S(T1) = \frac{SIF(T1)}{BRC\ SIF(T1)} \tag{7}$$

and

$$R_N = \frac{N}{BRC\ f_N} \tag{8}$$

These two parameters are plotted in Figure 11 for all of the ejection simulations with the symmetrically located (with respect to the X-Z plane) point masses. The point mass locations are arranged in order of increasing distance forward from the head c.g. or, for locations 3 plus 5 and 4, in the order of increasing radial distance. One conclusion which can immediately be drawn from this Figure is that the inertial loading effects of HMD become increasingly more severe with increasing distance of the HMD c.g. forward from the head c.g.

The results plotted in Figure 11 appear to be ideally suited for interpretation in a pass/fail sense. Thus if the pass criteria are taken to be both R_N and $R_S(T1) \leq 1.0$, the

cases which pass are AGH (generic helmet only), CG1 and 2, ASPS1 and S1. None of the cases considered at locations 2 and 3, which are common attachment sites for HMD such as CBO masks, night vision imaging systems and visors, pass. Note that all of the simulations included the effects of a 1 kg generic helmet with I_y = 100 kg-cm^2. Thus, according to Figure 11, the upper bound on the mass of a head mounted ensemble (i.e., helmet + mask + additional HMD) is 3 kg, provided the c.g. of the ensemble is coincident with that of the head and I_y of the ensemble does not exceed 100 kg-cm^2. It is unlikely that I_y of a 3 kg mass HMD ensemble would be less than 100 kg-cm^2 -- a relatively light helmet plus mask combination, such as the HGU-55/P plus the MBU-12/P, which weighs approximately 1.4 kg, has an I_y in excess of 100 kg-cm^2. Thus the maximum allowable mass for an HMD ensemble with c.g. coincident with the head c.g. appears to be less than 3 kg and the maximum allowable HMD mass above the 1 kg generic helmet appears to be less than 2 kg.

The HMD mass limit decreases with increasing distance from the head c.g. (particularly anteriorly). For an HMD with c.g. at location 3 and with a "counterweight" at location 5, the mass limit as indicated in Figure 11 is approximately 1.1 - 0.55 = 0.55 kg. The "counterweight" mass is subtracted off since it is merely a dead weight added to the ensemble to reduce the potential for neck muscle fatigue. It has nothing to do with the actual operation of the HMD. For location 4, the HMD mass limit appears to be approximately 1 kg; and for locations 3 and 2, approximately 0.6 kg.

When one considers that a typical helmet plus mask ensemble -- worn in the high speed, fixed-wing aircraft operational environment, which, in an emergency, can require crewmember ejection -- can weigh approximately 2 kg, the HMD mass limits indicated in Figure 11 appear to be somewhat conservative. A likely source for this conservatism comes from the following. The half-sine acceleration profile used to establish the BRC was identified as a moderate risk exposure. It should be emphasized, however, that it is a moderate risk exposure for the lower thoracic and lumbar spines. Based on AAMRL compilations of ejection acceleration induced spinal injury statistics, the likelihood of cervical spine vertebral body compressive fractures during ejection acceleration exposure appears to be significantly lower than the likelihood of vertebral body compressive fractures in the lower thoracic and lumbar spines. Thus, while the BRC for the lower thoracic and lumbar spines may indeed represent moderate risk criteria, the BRC used in generating Figure 11, i.e., the BRC f_N and SIF(T1) could very well represent low risk criteria. Since our goal was to establish guidelines based on moderate risk criteria, the results given in Figure 11 are probably conservative.

Analytical Evaluation of Specific HMD

The inertial loading effects of five helmet plus mask combinations were evaluated by comparing the R_N and R_S(T1) computed for the ejection simulations with those ensembles to the preliminary guidelines contained in Figure 11. The first combination, HGU-55/P + MBU-12/P, represents a standard pilot's configuration. The remaining four combinations; HGU-55/P + MBU-13/P, HGU-55/P + AR-5, HGU-39/P + MBU-13/P, and HGU-39/P + AR-5, represent four possible CD configurations.

The R_N and R_S(T1) for the ejection simulations with the specific HMD are listed in Table 3 along with the same parameters for simulations AGH, CG1, CG2 and CG3. All the R_N and R_S(T1) for the specific HMD configurations are less than 1.0. Thus all of these configurations pass the criteria that both R_N and R_S(T1) be $\leq$1.0. While the R_N and R_S(T1) appear to vary nearly linearly with mass for the generic HMD, their variations with mass and I_y of the specific HMD configurations are considerably less linear. This occurs because, while the c.g. for the generic HMD at location 1 (the head c.g.) is constant, the c.g.s for the specific HMD vary as indicated in Table 3. It is quite evident from Figure 11, that HMD c.g. location can be as significant with regards to HMD inertial loading effects as mass or moment of inertia.

CONCLUSIONS

The following are the main findings of the analytical investigation.

1. The inertial loading effects of HMD are observable in the internal loads developed in the neck and throughout the TL spine with the severity of these effects increasing with increasing (towards the head) spinal level.

2. Two parameters particularly useful in evaluating the inertial loading effects of HMD are R_N, the ratio of the computed neck injury parameter (NIP) to the baseline response criteria (BRC) NIP; and R_S(T1), the ratio of the computed SIF(T1) to the BRC SIF(T1).

3. The results contained in Figure 11 represent preliminary guidelines for limiting HMD mass and location with respect to the head c.g. for the purpose of minimizing the inertial loading effects of such devices during ejection acceleration exposures. The HMD pass criteria contained in these guidelines are R_N and R_S(T1)$\leq$ 1.0. These criteria appear to be conservative when viewed as moderate risk (5% probability of injury) criteria.

4. The inertial loading effects of HMD become increasingly more severe as they are located increasingly further, particularly anteriorly, from the head c.g.

5. All of the specific HMD ensembles -- helmet + mask combinations -- considered satisfy the pass criteria, R_N and R_S(T1)$\leq$1.0.

6. For the four CD configurations: the two involving the HGU-55/P helmet are less severe in terms of their inertial loading effects than the two involving the HGU 39/P.

The flexible rubber shroud of the AR-5 posed significant difficulties during the inertial properties measurements. The shroud had to be rolled/folded together and lumped at the base of the helmet so that the measurement procedure could be executed. As the shroud is actually at least partially draped over the air person's shoulders, the coupling of the entire shroud to the base of the helmet most likely compromised our measurements. Since these data were used directly for the HSM simulations, we feel that it is not appropriate to use the results listed in Table 3 to quantitatively compare the inertial loading effects of the AR-5 and MBU-13/F CBO masks.

The analytical investigation described in this paper and the related experimental work discussed in (3) and (5), have demonstrated analytical and experimental methodologies required to 1) establish general HMD design guidelines, and 2) define the inertial properties and evaluate the inertial loading effects of specific existing and planned HMD ensembles. This effort also produced some HMD design guidelines for ejection acceleration exposures. Based on the results obtained and the experience gained from this program, we have defined further analytical and experimental investigations designed to produce 1) general HMD design guidelines for various acceleration environments in the form of, e.g., spatial envelopes of HMD mass limits versus the coordinates of HMD c.g.s, and 2) accurate measurements of the inertial properties and evaluations of the inertial loading severities associated with specific existing or planned HMD ensembles.

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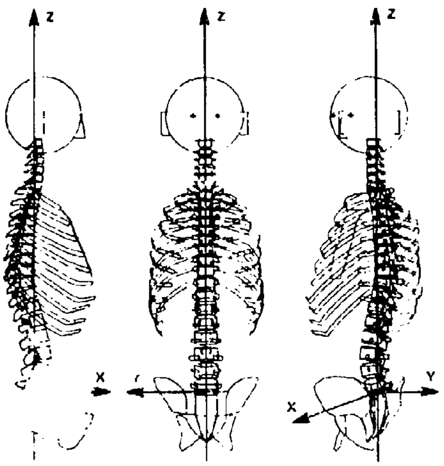

Figure 1. MID-SAGITTAL (X, Z), FRONTAL (Y, Z) AND OBLIQUE VIEWS OF HSM GEOMETRY

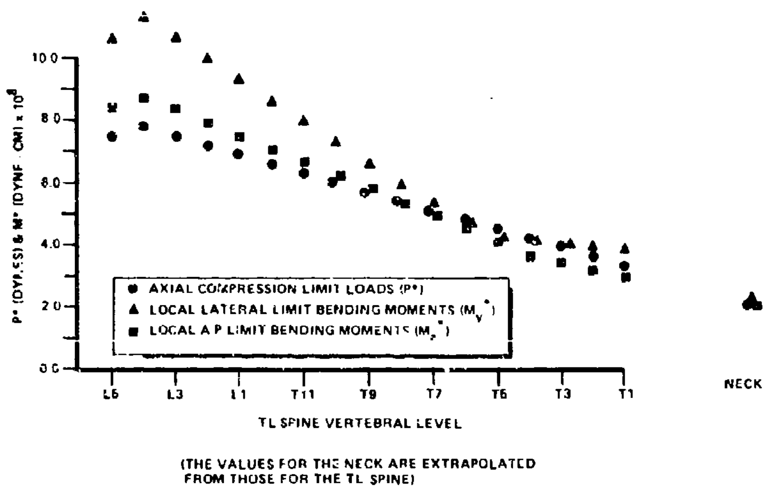

Figure 2 LIMIT LOAD DISTRIBUTION FOR THE TL SPINE (NECK VALUES ARE EXTRAPOLATED FROM THOSE FOR THE TL SPINE)

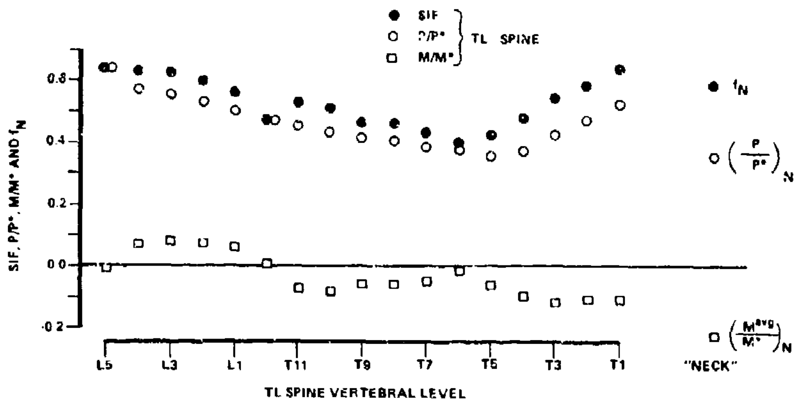

Figure 3 SIF, P/P*, M/M* AND f_N FROM BASELINE SIMULATION-FILLED IN CIRCLES ARE BRC

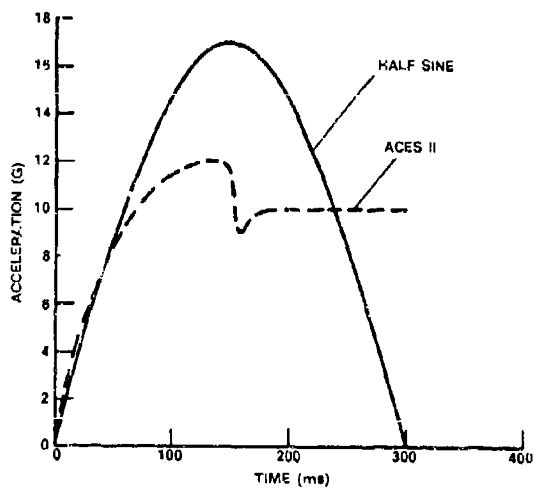

Figure 4 17G PEAK,300 ms HALF-SINE AND NONIMAL ACES II CATAPULT PLUS ROCKET ACCELERATION PROFILES

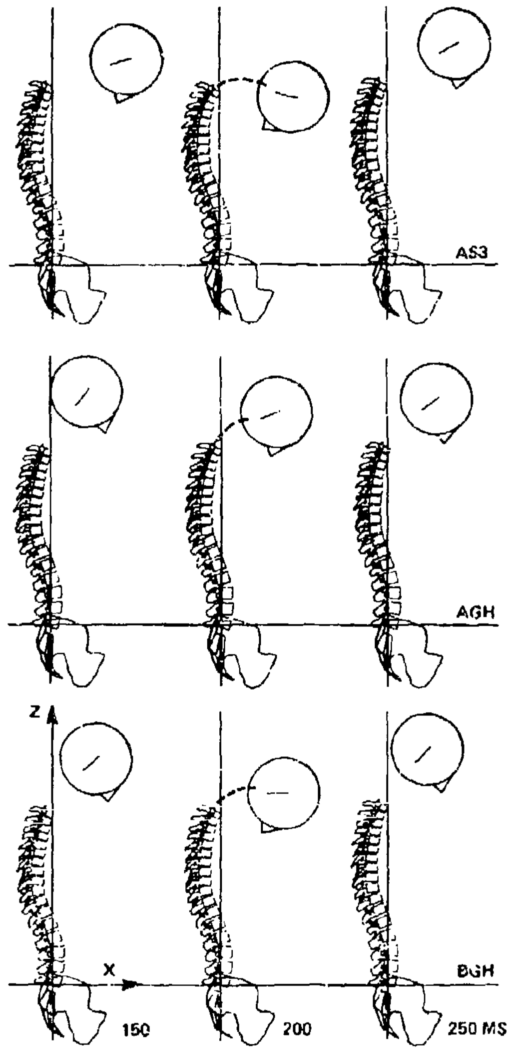

Figure 5 HSM MID-SAGITTAL (X,Z) PLANE CONFIGURATIONS AT 150, 200 AND 250 MS FOR SIMULATIONS BGH, AGH AND AS3

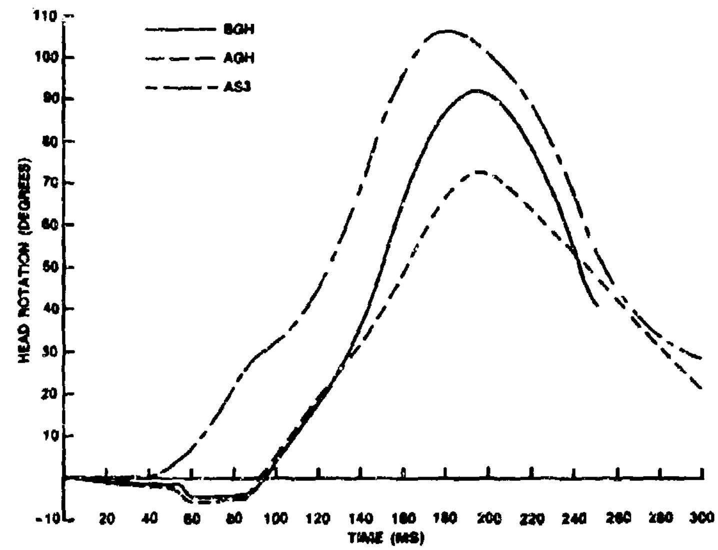

Figure 6 MID-SAGITTAL (XZ) PLANE HEAD ROTATIONS FROM SIMULATIONS BGH, AGH AND AS3

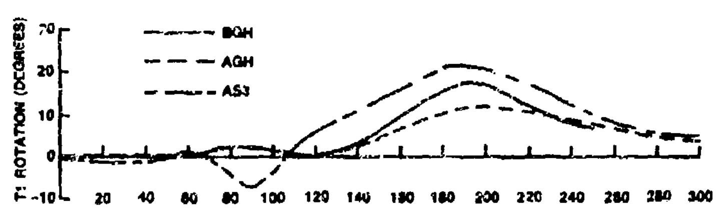

Figure 7 MID-SAGITTAL (XZ) PLANE T1 ROTATIONS FROM SIMULATIONS BGH, AGH AND AS3

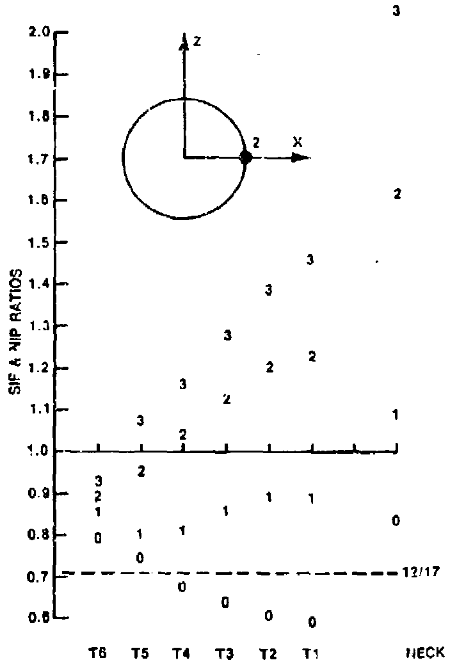

Figure 8 SIF RATIOS AND NIP RATIO FOR SIMULATIONS AGH, A1, A2 AND A3: 0, 1, 2 AND 3KG POINT MASSES AT LOCATION 2, THE ANTERIOR POINT

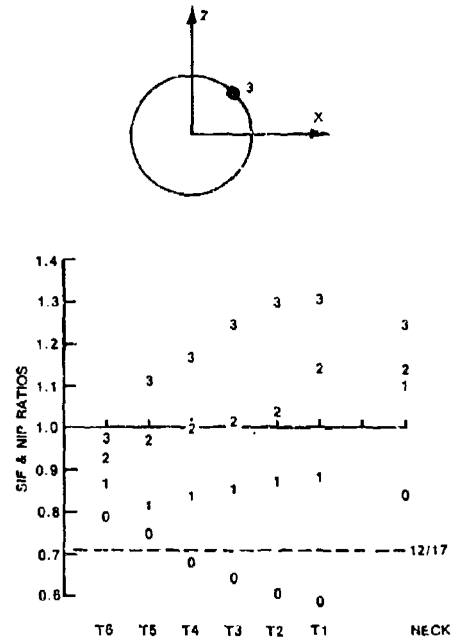

Figure 9 SIF RATIOS AND NIP RATIO FOR SIMULATIONS AGH, AS1, AS2 AND AS3: 0, 1, 2 AND 3KG POINT MASSES AT LOCATION 3, THE ANTERIOR-SUPERIOR POINT

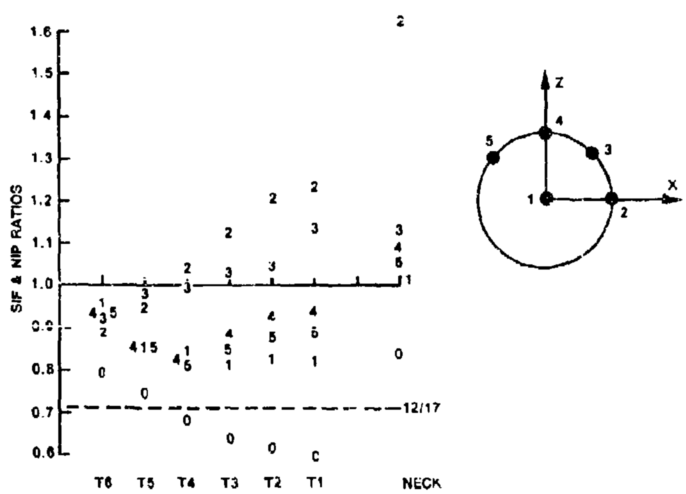

Figure 10 SIF RATIOS AND NIP RATIO VERSUS LOCATION OF 2KG POINT MASS

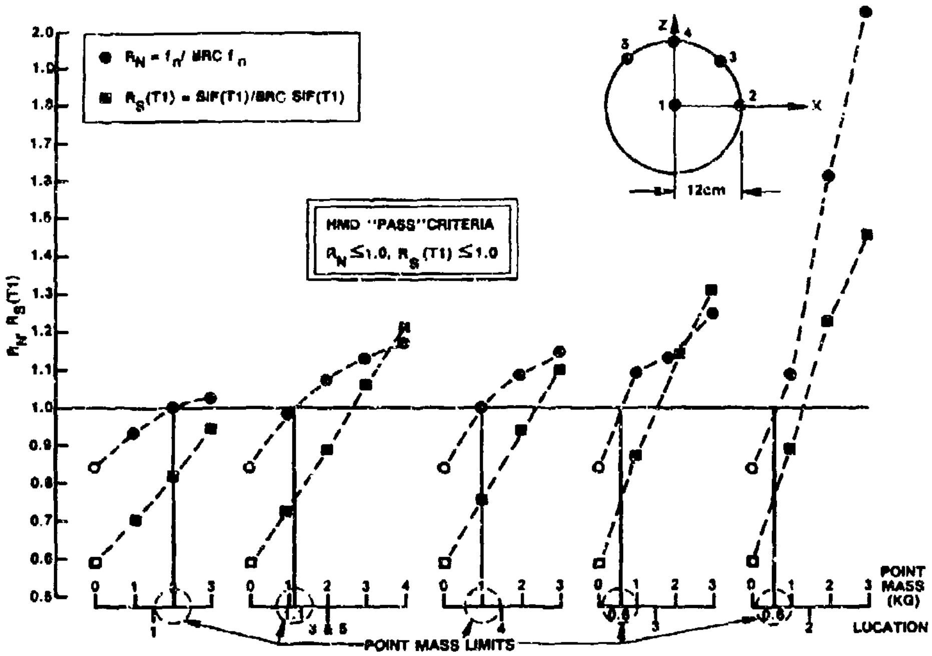

Figure 11 R_N AND R_S(T1) FOR ALL HSM EJECTION SIMULATIONS WITH SYMMETRICALLY LOCATED POINT MASS: POINT MASS LIMITS

MODIFICATION DE LA DYNAMIQUE DE LA TETE, CHARGEE PAR DES MASSES ADDITIONNELLES

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RESUME

De nombreux dispositifs imposent à leurs utilisateurs le port de masses rapportées sur la tête. Leur implantation obéit essentiellement à des impératifs techniques. Il est rare que les répercussions possibles sur le maintien et le mouvement de la tête soient envisagées au moment de leur conception.

Nous avons étudié cette question pour un système effectivement utilisé. Dans un premier temps nous avons déterminé la position du centre d'inertie et les moments d'inertie principaux du système considéré par une méthode expérimentale de pendulation. Nous avons ensuite précisé les modifications des paramètres inertiels de l'ensemble "tête + système additionnel" en analysant le déplacement du centre d'inertie et les variations des moments principaux par rapport à la tête nue, en relation avec les centres de rotation qui eux restent inchangés.

L'influence des masses additionnelles a été observée sur quatre sujets pour un mouvement calibré et facilement identifiable, défini par l'acquisition visuelle d'une cible imposant des rotations de la tête dans diverses directions, avec ou sans masses additionnelles. En ce qui concerne strictement l'aspect dynamique, nous avons relevé des variations du niveau et de la durée des accélérations de rotation en relation avec l'augmentation de l'inertie ainsi créée. Le déport du centre d'inertie induit également des perturbations verticales de l'accélération. Ce fait est confirmé par une augmentation très nette des activités des muscles de maintien et de mise en mouvement de la tête, en particulier les muscles de la nuque.

Un autre fait au moins aussi important doit être signalé. En effet la perturbation créée par l'implantation de l'appareil concerne également le maintien postural. On observe en particulier de fortes variations du stabilogramme pendant les expériences d'acquisition de cible lorsque le sujet est en position debout.

Cette modification de la dynamique de la tête entraîne une sensible dégradation de la performance, évaluée en fonction des erreurs relevées et de l'accroissement des temps de réponse pour la localisation de cibles aériennes.

L'attention des concepteurs de matériels doit donc être attirée sur l'influence déterminante de l'efficacité des opérateurs des apports de masse sur la tête. A partir des constats établis dans cette étude, une recherche plus générale doit être entreprise avec pour objectif d'établir les fonctions de sensibilité d'un opérateur aux inerties additionnelles, en liaison avec des critères de performance.

1 - INTRODUCTION -

Le port du casque traditionnellement prévu comme moyen de protection de la tête est de plus en plus fréquemment envisagé comme une possibilité d'implantation d'équipements complémentaires. Ces équipements sont destinés soit à la fourniture d'informations externes au sujet, visuelles ou auditives, soit au contraire à renseigner le système extérieur sur l'état du sujet : position de la tête, orientation du regard, etc...

Quel que soit le soin apporté à la réalisation de ces dispositifs par le choix et la répartition des matériaux, ils sont à l'origine d'une modification sensible de la charge pondérale de la tête et leur implantation, nécessairement située sur la périphérie du casque, se traduit par une altération importante des caractéristiques inertielles de la tête.

Aux effets de fatigue, ou voire même lésionnels, occasionnés par le port de ces systèmes pour une longue durée, viennent s'ajouter des effets immédiats ou à très court terme sur la statique et la dynamique de la tête.

L'objet de cette présentation est de rendre compte des travaux préliminaires qui ont été effectués relativement à ces effets à court terme, lors de l'évaluation d'un système destiné à détecter les mouvements de tête au moyen d'inclinomètres intégrés à un ensemble coiffe et casque.

L'approche expérimentale a porté sur deux aspects complémentaires :

- l'évaluation des modifications apportées à la tête, concernant les centres d'inertie et moments d'inertie principaux,
- l'étude de la dégradation des performances pour des mouvements d'acquisition d'une image en champ lointain, tête libre et tête chargée.

2 - MODIFICATIONS INERTIELLES DE LA TETE -

La première phase de cette étude a été centrée sur la détermination des caractéristiques propres du système implanté sur la tête. A cet effet nous avons utilisé la méthode classique du double pendule. Le casque est incorporé à une nacelle pouvant successivement osciller autour de deux axes parallèles. De la mesure des deux périodes d'oscillation, associée à la connaissance de la masse de l'ensemble et de l'entraxe, on peut déduire aisément à la fois la position du centre d'inertie relativement à l'un ou l'autre axe d'oscillation et le moment d'inertie par rapport à ces directions. Les mêmes mesures étant faites sur la nacelle vide, conduisent par différence aux caractéristiques du casque seul. En procédant successivement selon les trois directions orthogonales de l'espace : verticale, antéro-postérieure, latérale, on a obtenu dans un référentiel lié au casque la position du centre d'inertie et les trois moments d'inertie centraux (Tableau n°1).

Ces informations doivent ensuite être associées aux caractéristiques propres à celles de la tête, centre d'inertie et moments d'inertie. Ces données sont généralement obtenues à partir de relevés photogrammétriques permettant de déterminer les coordonnées de points situés sur des courbes de niveau et de calculer ensuite, par intégration numérique approchée, les différentes caractéristiques inertielles pour une masse volumique uniforme voisine de l'unité. De tels travaux ont été réalisés dans notre laboratoire (R. MOLLARD, 1987), mais ont été également publiés par McCONVILLE J. et Coll. (1980). Ces travaux ont permis d'établir des méthodes d'estimation des caractéristiques inertielles segmentaires pour les différents éléments anatomiques, en utilisant comme données de base des mesures anthropométriques classiques.

Ce principe a été retenu pour estimer les caractéristiques inertielles de la tête des quatre opérateurs ayant participé aux évaluations de performance (Tableau n°2). Les mêmes évaluations ont été réalisées pour une population générale de militaires français extraite de ERGODATA (Tableau n°3) et on peut constater que les sujets retenus pour les tests de performance présentent une variabilité interindividuelle très importante pour ces caractéristiques inertielles, couvrant ainsi l'amplitude totale de variation de la population de référence.

On peut ensuite grâce à des relations classiques de la cinétique, caractériser l'ensemble tête casque. Selon les indications du schéma correspondant au plan sagittal médian, on constate un déplacement très important du centre d'inertie de la tête lorsque le sujet est équipé de son casque, de l'ordre de 4 cm. (Figure n°1). Par ailleurs la variation des moments d'inertie est, comme l'indique le tableau n°4, très importante. Ce résultat est d'ailleurs prévisible puisque ici les matériaux additifs ont une masse volumique moyenne beaucoup plus élevée que celle de la tête, et sont implantés loin des centres. Ici nous avons ramené les valeurs des inerties aux axes de rotation de la tête, puisque ce sont elles qui interviennent significativement dans les mouvements.

3 - EVALUATION DE L'ENSEMBLE HOMME-EQUIPEMENT -

Le second objectif de cette étude préliminaire était de caractériser la réponse dynamique de la tête avec ou sans casque, lorsque le sujet doit porter son attention visuelle sur un point de son champ soudainement occupé par une image.

La méthode repose sur des présentations de tâches élémentaires destinées à recréer pour le sujet des conditions d'observation, d'acquisition et de suivi de cibles aériennes, au moyen de projections d'images sur un figuratif de ciel, quart de sphère de 10,4 m. de diamètre (Figure n°2). L'apparition de la cible est aléatoire, et le sujet en position initiale neutre doit la localiser et informer le système par action sur un bouton-poussoir ; la sollicitation est ou séquentielle, ou insérée dans une phase de suivi de cible. L'ensemble de l'expérience est gérée par un logiciel spécifique, permettant de quantifier les tests de performance sous forme de mesure de temps de réponse, de taux d'erreurs de localisation et d'identification de la cible.

Parallèlement l'activité du sujet est contrôlée par un ensemble de mesures biomécaniques :
- mesure des rotations droite-gauche, à l'aide d'un potentiomètre placé sur la tête du sujet,
- accélérations tri-axiales captées au niveau du front,
- posturographie, au moyen d'un statokinésimètre placé sous les pieds de l'opérateur qui permet l'enregistrement des efforts transmis à l'interface pied-sol, suivant deux axes orthogonaux de référence, antéro-postérieur et latéral,
- électrocardiographie et électromyographie des groupes musculaires suivants :
  - cou : sterno-cleido-mastoïdien,
  - nuque : trapèzes moyens, droit et gauche,
  - dos : spinaux côté droit.

Nous avons testé les quatre sujets en retenant pour chacun d'eux comme éléments de référence, les résultats des essais tête libre. Les essais se sont déroulés selon divers protocoles.

Le premier consiste dans l'acquisition de cibles nécessitant des rotations droite et gauche de la tête à partir d'une position neutre centrale, soit 20 cibles pendant cinq minutes.

Ensuite on traite, toujours en cinq minutes, 20 cibles à apparition aléatoire localisées en positions :
- droite-haute,
- droite-basse,
- avant-haute,
- gauche-haute,
- gauche-basse.

Enfin pendant des durées plus longues (15 à 30 minutes) il s'agit de suivis de cibles avec acquisitions : les trajectoires sont circulaires horizontales, à plusieurs niveaux, ou sinusoïdales avec ou sans phases d'arrêt et des vitesses de déplacement constantes ou variables. Au cours de ces suivis l'acquisition consiste en l'extinction de la cible suivie de l'apparition de manière aléatoire d'une autre cible de figuratif de ciel, dans une autre position de référence (20 ou 40 cibles).

Les résultats présentés sur les tableaux n°5 à 8 font apparaître une dégradation très sensible des performances des sujets, accompagnée d'une augmentation très nette de l'amplitude des stabilogrammes et des activités E.M.G. des muscles sollicités lors des mouvements de la tête (Figures n°3 à 5).

Les accélérations tangentielles et normales sont diminuées en amplitude de manière notable tandis que les durées de signal augmentent, reflétant ainsi directement l'effet de l'inertie (Figure n°6). Les accélérations verticales sont plus élevées. Les temps de réponse sont en augmentation de 15 à 40% avec une augmentation de la dispersion de leur valeur, tandis que les forces transmises au sol dans le sens antéro-postérieur sont en augmentation de 10 à 15%, ce qui apparaît à l'examen du stabilogramme (cf. figures n°3 à 5).

4 - CONCLUSIONS -

Les variations très importantes des caractéristiques de la tête causées par le port du système modifient le comportement dynamique de la tête. Le déport du centre d'inertie a pour effet de déstabiliser l'asservissement de

maintien de la tête, ce qui est également traduit par l'augmentation de l'activité des muscles impliqués dans un mouvement. Il est important d'en souligner l'influence corrélative sur le maintien et la régulation posturale et le résultat final quant à la dégradation de la performance. Il y a cependant une certaine adaptation au système puisque les perturbations observées ne sont pas accrues après une heure de port du casque. A l'évidence le cas présenté ici est un cas limite, et il est clair qu'on doit s'attacher à rester dans un devis de masse nettement inférieur.

Cependant on peut retenir que les tests choisis sont significatifs de la perturbation étudiée. Ils peuvent servir d'outils de base pour entreprendre une étude expérimentale beaucoup plus vaste qui consisterait en une cartographie de l'influence de charges sur la tête, en termes de résultantes et de moments résultants appliqués en différents points. En effet une telle démarche permet de faire varier séparément les effets de la charge proprement dite et les effets de son moment, combinaison de la charge et des distances de son implantation au centre de rotation, un couple donné pouvant à l'évidence être induit par différentes répartitions de charge. On peut espérer définir à partir d'une telle étude, des critères d'optimisation utiles aux concepteurs d'équipements.

Masse ≈3Kg	Valeurs des moments d'inertie du casque (exprimées en $kg.m^2$)
I_x	$2,53.10^{-2} \pm 0,01$
I_y	$2,91.10^{-2} \pm 0,01$
I_z	$3,12.10^{-2} \pm 0,02$

Tableau n°1
Valeurs des inerties principales du casque.

SUJETS	S1	S2	S3	S4	Erreur d'estimation au seuil de probabilité $P = .05$
Longueur de la tête (cm)	19,1	19,1	19,3	19,6	
Périmètre de la tête (cm)	54,0	56,0	58,2	60,0	
I_x $(kg.m^2)$	$1,64.10^{-9}$	$1,87.10^{-9}$	$2,14.10^{-9}$	$2,41.10^{-9}$	$\pm 0,32.10^{-9}$
I_y $(kg.m^2)$	$1,74.10^{-9}$	$2,10.10^{-9}$	$2,45.10^{-9}$	$2,81.10^{-9}$	$\pm 0,37.10^{-9}$
I_z $(kg.m^2)$	$1,18.10^{-9}$	$1,41.10^{-9}$	$1,63.10^{-9}$	$1,85.10^{-9}$	$\pm 0,18.10^{-9}$

Tableau n°2
Estimation des moments d'inertie Ix, Iy, Iz de la tête des sujets ayant participé aux expérimentations.

	Moyennes et écart-types des mesures de référence et amplitudes de variations des valeurs inertielles		Erreur d'estimation au seuil de probabilité $P = .05$
Longueur de la tête (cm)	$m = 19,31$	$\sigma = 0,66$	
Périmètre de la tête (cm) .	$m = 56,45$	$\sigma = 1,54$	
$I_{x\,(kg.m^2)}$	$1,5.10^{-9} \leqslant I_x \leqslant 2,34.10^{-9}$		$\pm 0,32.10^{-9}$
$I_{y\,(kg.m^2)}$	$1,63.10^{-9} \leqslant I_y \leqslant 2,72.10^{-9}$		$\pm 0,37.10^{-9}$
$I_{z\,(kg.m^2)}$	$1,1.10^{-9} \leqslant I_z \leqslant 1,9.10^{-9}$		$\pm 0,18.10^{-9}$

Tableau n°3
Estimations et amplitudes de variations les plus probables de valeurs inertielles d'une population de jeunes français du sexe masculin (N = 794) dont on connaît les paramètres statistiques de mesures anthropométriques relevées sur la tête.

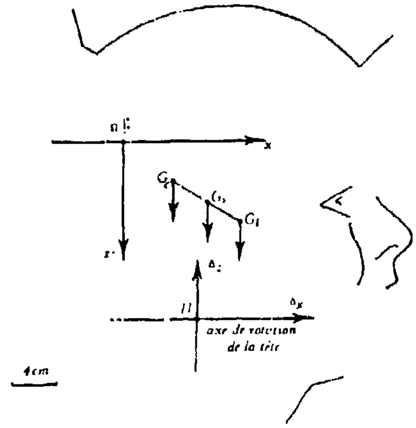


Figure n°1
Position du centre de gravité Gr résultant de l'ensemble tête-coiffe dans le plan sagittal médian et direction des axes de rotation de la tête.

Sujet *S4* - Moments d'inertie (kg. m^2)	
Tête ($I_{\Delta_{1/Y'Y}}$)	$3,31.10^{-2}$
Casque ($I_{\Delta_{2/Y'Y}}$)	$4,1\ .10^{-2}$
Tête + casque ($I_{\Delta_{/Y'Y}}$)	$7,41.10^{-2}$
Tête ($I_{\Delta_{1/Z'Z}}$)	$3,14.10^{-3}$
Casque ($I_{\Delta_{2/Z'Z}}$)	$3,19.10^{-2}$
Tête + casque ($I_{\Delta/Z'Z}$)...............	$3,5\ .10^{-2}$

Tableau n°4
Valeurs des moments d'inertie de la tête, du casque et de l'ensemble (tête-casque) du sujet S4 par rapport à l'axe de rotation de la tête - pour deux directions perpendiculaires : Y'Y, Z'Z.

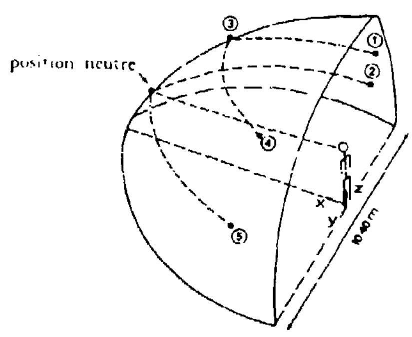


Figure n°2
Schéma représentant les différentes localisations des cibles pour les phases d'acquisition.

PROTOCOLE n°1 - Temps d'acquisition Droite/Gauche

SUJETS	A Tête nue		B Tête + CASQUE		B - A en %
	m	σ	m	σ	
S1	936	103	1066	116	+14
S2	644	53	829	95	+29
S3	756	167	1082	705	+43

Tableau n°5
Comparaison des temps de réponse pour des rotations droite/gauche de la tête, avec ou sans port du casque.
(m : moyenne et σ : écart-type - Valeurs en millisecondes).

SUJETS		2 + 5	2 (Droite)	5 (Gauche)
S1	A	936	939	935
	B	1066	1057	1078
S1		644	622	693
		829	841	813
S2		756	817	709
		1082	1270	873

Tableau n°6
Variation des temps moyens d'acquisition en fonction de la localisation spatiale de la cible.
(Valeurs en millisecondes).

PROTOCOLE n°2 - Acquisition avec ou sans suivi de cible

SUJETS	A Tête nue		B Tête + Casque		B - A %	A Tête nue		B Tête + Casque		B - A %
	m	σ	m	σ		m	σ	m	σ	
S1	939	159	1780	802	+89	1005	396	1910	1006	+90
S2	1129	306	1355	245	+20	1282	670	2040	983	+59
S3	1252	99	2043	287	+63	1351	501	1914	845	+42
S3	1243	165	1919	195	+54	1629	727	1804	909	+11
S4	1277	378	1552	265	+21	1604	752	1850	494	+15

Tableau n°7
Comparaison des temps de réponse pour différentes tâches d'acquisition avec ou sans port du casque.
A = Condition de référence : tête nue et B = Tête + casque.
(Valeurs en millisecondes).

PROTOCOLE n°2 - Tâches d'acquisition

SUJETS	GLOBAL	1 droite haut	2 droite bas	3 avant haut	4 gauche haut	5 gauche bas
S1	(A) 939	824	896	814	1023	1156
	(B) *1780*	*2867*	*1405*	*1216*	*2296*	*1427*
S2	1129	1012	1156	942	1249	1301
	2366	*1273*	*1263*	*1306*	*1440*	*1453*
S3	1252	1215	1280	1214	1229	1373
	2043	*2224*	*1846*	*1997*	*2209*	*1657*
S3	1243	1292	1244	1171	1301	1157
	1919	*1895*	*1736*	*1656*	*2026*	*1634*
S4	1277	1185	1099	1149	1379	1562
	1552	*1571*	*1331*	*1464*	*1881*	*1634*

Tableau n°8
Variations des temps d'acquisition en fonction de la localisation spatiale de la cible.
A - Condition tête nue - référence et B = Tête + Casque (Valeurs en millisecondes).

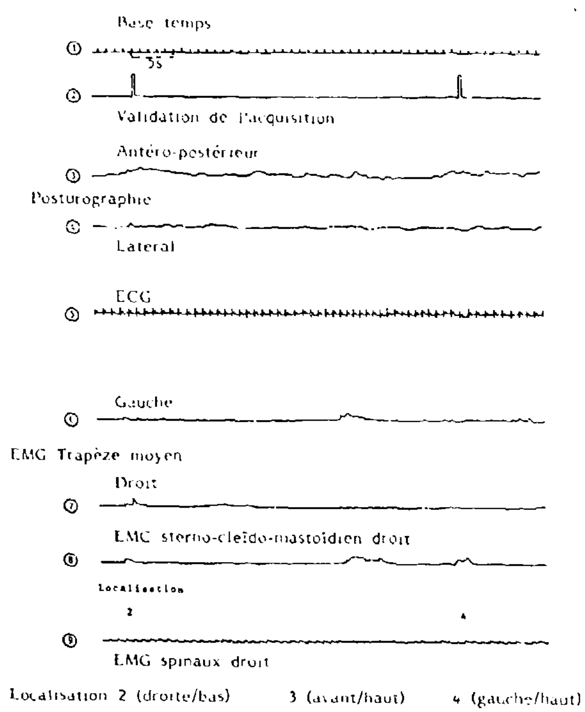


Figure n°3
Exemple d'enregistrement de données biomécaniques et physiologiques pour le Sujet S3.
Conditions d'essais : sans casque ; acquisition de cibles pendant suivis - Test 2.

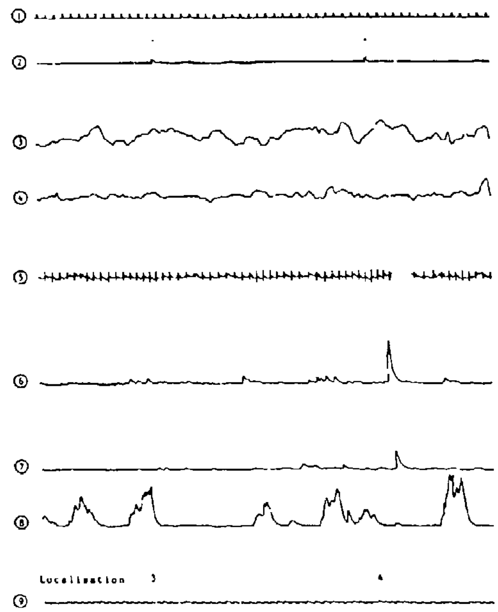


Figure n°4
Exemple d'enregistrement de données biomécaniques et physiologiques pour le Sujet S3.
Conditions d'essais : avec casque ; acquisition de cibles pendant suivis - Test 2.

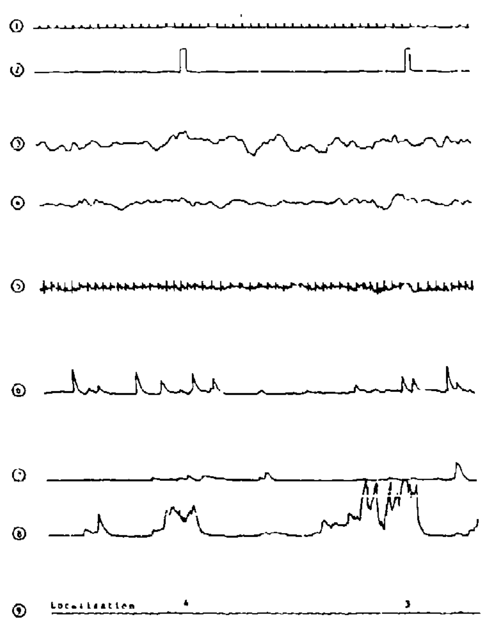
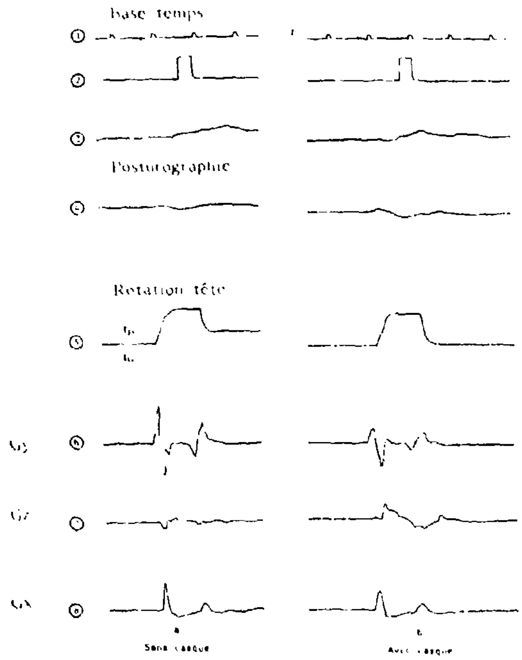


Figure n°5
Exemple d'enregistrement de données biomécaniques et physiologiques pour le Sujet S3.
Conditions d'essais : avec casque (après 2h. de port) ; acquisition de cibles pendant suivis - Test 2.



<u>Figure n°6</u>
Exemple d'enregistrement pour un mouvement de rotation de la tête vers la droite, en condition tête nue (a), puis avec le port du casque (b) - Sujet S1.
Conditions d'essais : acquisition horizontale.

5 <u>REFERENCES</u> -

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DISCUSSION PERIOD 3

Col Asleben, GAF.

Ihave a question for Professor Snijders or any of the other speakers this morning. Working on EFA (the European Fighter Aircraft) program and I am responsible on the German side for operational aspects. We need a helmet with helmet mounted devices. As you know EFA will be more agile than the F16 with higher G onset rates.:

1)What would you consider would be the maximum weight that the pilot can stand long term without getting fatigue and in the short term without getting injured ?

2)What movement or displacement of the C of G fulfills the same as another weight which increases the total mass but equalises the balance.?

Professor Snijders, Netherlands

We need very much more information such as was said on one of this mornings papers on the failure analysis of vertebrae and soft tissues. According to our results we are near these limits, so to increase the weight of even helmet and helmet mounted devices will be tricky. We cannot predict the long term effects on the spine therefore we need more epidemiology and X-ray studies. What I want to emphasize is already at this time it is a pity that the helmet adds so much weight in a situation where you want to get rid of it. To decrease the weight certainly by 30-50% would be a really important influence on the weight of the neck and that also would decrease muscle fatigue dramatically. According to our figures within 10-30secs some operation like air combat would lead to absolute fatigue. In other positions with lower G values such as adjusting the computer, other problems exist. In general, in answer to your question, I would say we can improve a little bit by changing the mass centre of gravity of the helmet.

Col Asleben, GAF.

This does not answer my question. We have to give an answer to the engineers right now and say the maximum weight should be, say 1.4-1.5Kg, because they have to start the work on this for future development. So what would you consider should be the maximum weight.

Professor Snijders, Netherlands.

The helmet in our study was the Alpha helmet. If you take that as an example I would say that was the limit.

Dr Privitzer, USAF.

I can't really speak for the Air Force anymore, but the approach we took is to look at the system you have in mind, come up with the inertial properties and run simulations in which we expose the head spine model with those inertial properties incorporated, to the types of environment that you are concerned with, and look at what the mass limits were. According to our criteria we want certain response parameters to stay below certain levels. The type of levels I was talking about correspond to yield stress on the neck and cervical spine elements and the upper thoracic level. There is a question of how valid it is and there is still work to be done but we seem to be in a reasonable ball park. You want a number but right now you can tell from the presentations you can't do this yet.

Mr Frisch, USA.

The Navy is in the process of an introducing an integrated night vision system into attack aircraft. The question is one that we have wrestled with for some time. The approach that we have taken is that we know that many of the helmet types we have flown in the past have been much heavier than ones flown now. So as far as weight is concerned we can probably give a little territory. From all the studies we have conducted in terms of head and neck response in mannekins and mathematical simulation, the crucial parameter seems not to be so much the mass but the C of G location of the system. So what we went out with is using the HGU 33P as a baseline. We wanted the three manufacturers that bid on the system to come up with something that weighs the equivalent of the HGU 33P and have a more favourable C of G location closer to the occipital condyles. Nobody I think will be able to tell you what the absolute weight is.

Dr Leger, France.

I would like to put a question to Professor Snijders about the accuracy of the measurement used to establish your model, and also about the accuracy of inflight measurements of head displacement.

Professor Snijders, Netherlands.

There was not time in my presentation so I did not include all the details of the accuracies. They are of paramount importance. The measurements of the position of the head have been performed by the National Aerospace Laboratory in the Netherlands. They claim with their computing system that the positions could be determined with a SD of 1.5^{o} and the orientation of the direction of accleration which was obtained from accelerometers on the helmet based on piezoresistive transducers were of the same order of magnitude. What was very important in our study was the neutral position, because all calculations were started with that the neutral F16 position. We can say that 3^{o} may be inaccurate. Next we did a parameter deviation study (sensitivity analysis) there were so many parameters involved it came out that the model was most sensitive, as could be predicted, to the points of attachments of muscles, with respect to the axis of rotation of the different levels. By changing all parameters plus or minus 10% we found that there were deviations on the forces at the lower cervical level deviated up to 60%, so we have to be very careful with our interpretations. However +/- 10% we think is a little much and therefore we state again and again that the model is most appropriate for comparing different situations. By simulating an optimal helmet I have shown a theoretical helmet weight of 800g with the C of G as much behind the axis of rotation as now it is in front. Then you can obtain interesting improvements.

Dr Von Gierke, USA.

Dr McElhaney, you showed the parameter of the vertebra under various loads in a stabilised condition and in an equilibrated condition. I wonder going from the one to the other is some fluid being pressed out of the vertebrae which again is absorbed by the vertebrae by the time you insert them in saline. The question is, is the static height of the vertebrae different in the conditions and which condition is really closer to the in vivo situation of the vertebrae.

Dr McElhaney, USA.

The standing height definitely does change under those conditions, as far as which is closest to in vivo I think that it is well known when you get up in the morning you are at your maximum height and during the day you shrink as much as .5" and then you lay down and re-equiliberate your vertebrae. I think in vivo may be we are operating between these two extremes. There is definitely fluid exuded from the discs, more than anything else in the cyclic loading, and as far as potential for injury there is a theory in orthopaedics that you are more liable to injure your lumbar spine in the am than the pm when the discs are fully equilibrated and stiffer.

Dr Von Gierke, USA.

Well we repeated some of the classic movements that you have just quoted that you are shorter in the evening than you are in the morning. This might be true but the main effect comes from changing your posture and muscular tension throughout the evening and not so much from compressing the vertebrae during the day but the differences you have between the two conditions are very marked. There are quite substantial differences in the parameters of the vertebrae; the question really is, is some of this being caused by fatigue and compression of the vertebrae or is some of it just an artefact in the type of in vitro measurements that we do, where you cannot prevent pressing some fluid out of the vertebrae which is to some extent prevented in vivo.

Dr McElhaney, USA.

I think that, that is correct that there may be some artefact in our preparation. We do keep the specimens moist in 100% humidity during this testing. We have also found that we can go from one state, the fully equilibrated state, to the mechanically stabilised state and then back again so it is difficult for me to say how much this compares to the in vivo state but is a repeatable end state in the in vitro experiments that we do.

Dr Kriebel, GAF

If I got it right, none of your football players had a fracture of the dens axis, this is suprising. Did you find any fractures of the dens axis? The second question is you had quite a number of vertebral fractures. How many of these patients suffered symptoms of spinal cord compression and how many of the sportsmen could go back to sport again.

Dr McElhaney, USA.

The numbers that I mentioned on both the football and diving accidents were all permanently quadraplegic, so none recovered. We have not seen any fractures of the dens in these activities. The primary type of fracture in 75% or more is a compression fracture with anterior dislocation, usually in C4,5,6 region.

MOBILITE DE LA TETE ET FACTEUR DE CHARGE: APPROCHE EXPERIMENTALE EN CENTRIFUGEUSE

par

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RESUME :

L'environnement des pilotes d'avion de combat moderne est particulièrement agressif pour le système tête-cou. L'introduction de systèmes optroniques montés sur le casque aggrave encore le problème des accélérations + Gz.

Des études en centrifugeuse ont été menées afin d'évaluer l'influence des paramètres d'environnement (inclinaison du siège, accélération + Gz en plateau ou en variation) sur la mobilité de la tête.

280 lancements ont été effectués sans rencontrer d'incidents traumatiques. Les résultats obtenus montrent que jusqu'à 5 G, les caractéristiques de déplacement de la tête sont peu modifiées. En revanche la variation du niveau d'accélération amène des perturbations de la stabilité.

Les sous-systèmes réflexes impliqués dans la stabilisation de la tête étant influencés par le contrôle volontaire, les lois de commande des avions de combat pourraient constituer un point d'intérêt dans l'étude de la physiopathologie du système tête-cou du pilote.

1. - INTRODUCTION

Depuis plusieurs années, on constate dans la communauté aéronautique un intérêt certain pour le retentissement des accélérations + Gz sur le segment céphalique. Le problème de la mobilité de la tête en combat aérien et des éventuelles conséquences traumatiques ne constitue cependant pas un phénomène entièrement nouveau.

Trois facteurs peuvent être principalement avancés à l'origine de cet intérêt :

- L'introduction des commandes de vol électriques, qui contribue à la très grande manoeuvrabilité des avions de combat moderne, mais aussi à l'installation rapide et brutale des accélérations + Gz.

- L'inclinaison des sièges, destinée à protéger contre les accélérations + Gz, mais qui prive la tête d'un appui.

- Enfin, l'introduction de dispositifs optroniques montés sur le casque, inéluctable pour les avions futurs, améliore la performance du couple homme-machine, mais impose des masses additionnelles sur la tête du pilote.

Ce dernier point présente lui-même un double aspect. Le premier est purement biomécanique et s'intéresse au risque traumatique introduit par l'équipement de tête. Le second est lié à l'emploi de ces systèmes sous facteur de charge et concerne essentiellement le contrôle moteur de la tête et du cou en présence de perturbations de force extérieurement appliquées.

Diverses approches de ce problème peuvent être proposées. L'enquête épidémiologique permet de préciser les circonstances de survenue et l'incidence réelle des phénomènes en situation opérationnelle (1, 2, 9). Les techniques de modélisation mathématique constituent également un apport intéressant pour ce qui concerne la biomécanique des traumatismes du segment tête-cou (6, 8).

Depuis plusieurs années, le Laboratoire de Médecine Aérospatiale du Centre d'Essais en Vol a entrepris un programme d'étude approfondi sur l'effet des accélérations + Gz sur la mobilité de la tête. Différents aspects, comme l'inclinaison du siège et l'emploi des viseurs de casque sous facteur de charge, ont ainsi pu être abordés lors d'études expérimentales en centrifugeuse.

2. - METHODES

Les essais ont été conduits sur la centrifugeuse du Laboratoire. La longueur du bras est de 6 mètres, avec une nacelle pendulaire amortie. Bien que dotée d'une commande manuelle, la reproductibilité des niveaux d'accélération et de leur taux de variation est bonne. Pour ce qui concerne ce dernier point les pentes de mise en accélération peuvent varier entre 0,1 G/s à 1,2 G/s en utilisant alors un système de catapulte.

La nacelle utilisée pour les expérimentations sur la mobilité de la tête permet d'embarquer des expérimentations encombrantes, avec une charge utile de 270 kg.

Un total de 280 lancements a été réalisé au cours de 4 campagnes d'essais. La campagne initiale avait pour objectif essentiel d'observer l'effet de l'inclinaison du siège sur la mobilité de la tête sous facteur de charge. Les campagnes suivantes ont été plus spécialement consacrées aux problèmes d'emploi du viseur de casque sous facteur de charge, dans le cadre du développement de l'avion tactique issu du démonstrateur "Rafale A".

2.1. - Mobilité de la tête et inclinaison du siège.

Le dispositif expérimental utilisé pour cette étude a été mis en place autour d'un siège inclinable utilisé lors d'études antérieures de validation du concept. Deux inclinaisons du dossier du siège ont été retenues : 20° et 45°. Dans ce dernier cas, deux conditions initiales étaient utilisées, appui sur le repose-tête ou maintien de la tête alignée avec le vecteur gravité.

Des cibles fixes étaient présentées sur un mode pseudo-aléatoire, selon trois excentricités en gisement + 45°, + 90°, + 135° et différentes localisations en site sur les méridiens précédemment définis.

Les variables considérées étaient le temps d'acquisition visuelle (T1) et le temps d'alignement (T2) au moyen d'un dispositif monté sur le casque. Le casque utilisé pour cette étude avait une masse approximative de 1400 g . Les temps étaient mesurés à partir d'un signal donné par le sujet au moyen d'un bouton-poussoir.

Six désignations successives sur chaque méridien à droite et à gauche étaient effectuées lors d'accélérations stabilisées en plateau à + 3, + 4 et + 5 Gz. Des essais ont également été conduits à 7 G en utilisant uniquement des cibles à 45 et 90°. Dans ce dernier cas le nombre de désignations était limité à 3.

2.2. - Emploi du viseur de casque sous accélérations + Gz. (5).

Le dispositif expérimental a été installé autour d'une maquette géométrique du démonstrateur Rafale A avec un siège incliné à 32°.

Ce dispositif comprenait :

- Un écran hémisphérique de 170 cm de diamètre.

- Un dispositif optique de renvoi à l'infini d'une cible fixe, remplacé ultérieurement par un système de projection à deux degrés de liberté utilisant un rayon Laser.

- Un viseur de casque électro-optique THOMSON C.S.F La précision de ce système est de l'ordre de 0,5°. Ce viseur était monté sur un casque GUENEAU 458 modifié; la masse totale du système était alors de 1280 g. Il permet de déterminer avec précision les trois angles de la visée (site, gisement, roulis) ainsi que la position de l'origine de la visée dans le référentiel cabine.

Les protocoles d'études ont initialement porté sur l'acquisition et la désignation de cibles renvoyées optiquement à l'infini. Une localisation à + 50° en site et 25° en gisement a été retenue pour une première série d'essais. La mise au point d'une procédure de correction de l'erreur de parallaxe a conduit à une étude complémentaire, effectuée dans des conditions proches de la première campagne, avec une cible stationnaire à + 30° en site et en gisement. Ces deux campagnes ont été poursuivies par une étude sur cibles mobiles avec différentes trajectoires d'amplitude et de caractéristiques variables.

Les essais ont été menés dans deux conditions :

- En présence d'accélérations + Gz stabilisées en plateau à + 3, + 4 et + 5 Gz.

- En présence d'une variation d'accélération à 0,6 G/s avec des accélérations terminales en plateau identiques aux précédentes.

3. - RESULTATS

Trois points sont à considérer pour les résultats de cette étude : l'incidence traumatique, l'effet de l'inclinaison du siège et les caractéristiques cinématiques du déplacement de la tête sous facteur de charge.

3.1. - Incidence traumatique.

L'incidence traumatique relevée au cours des essais est extrêmement faible. Un seul sujet a abandonné les essais sans les terminer, en raison d'un épisode aigu de cervicalgie, survenu à l'issue d'une série de lancements.

L'examen médical a permis de conclure à une origine musculaire et la guérison a été rapide. D'autres sujets ont signalé des phénomènes de tension musculaire au niveau du cou à l'issue de série de lancement. Toutefois ces phénomènes disparaissent généralement d'une manière spontanée et rapide. Dans l'ensemble, les différents protocoles utilisés ont été bien supportés par les sujets, avec seulement quelques incidents mineurs qui ne remettaient pas en cause la participation à l'expérimentation.

3.2. - Effets de l'inclinaison du siège.

Les figures 1 à 3 présentent les résultats obtenus pour les temps d'acquisition visuelle T1 et d'alignement T2, en fonction de l'excentricité de la cible et de la configuration du siège et de la tête. Les valeurs sont présentées pour les essais de référence à 1 G et pour les essais sous facteur de charge à 3 G et 5 G.

L'examen de ces données montre que dès 3 G, pour les cibles les plus excentrées, le temps d'alignement augmente considérablement avec le siège à 20° et lorsque la tête est maintenue droite sur le siège à 45°. Pour les essais à 5 G, l'apparente diminution des valeurs de T2 avec le siège à 20° est liée au fait que 4 sujets sur 6 n'ont pu effectuer la tâche d'alignement.

En dépit de la très grande variabilité inter-individuelle, l'analyse statistique des données montre que :

- L'alignement de la tête est obtenue plus rapidement dans la position siège 45° tête appuyée (45 Ti) par rapport aux deux autres configurations. Cette différence est significative au risque 1 %.

- Il existe également une différence significative entre les valeurs obtenues en référence 1 G par rapport aux essais à 3 G et 5 G. Par contre l'analyse ne montre pas de différence significative entre ces deux niveaux.

Bien que ce problème ne soit pas directement lié à la mobilité de la tête, la figure 4 permet de considérer les résultats obtenus à 7 G. La variable considérée est le pourcentage des essais ou l'alignement de la cible a été obtenu d'une manière correcte, toutes excentricités confondues. A 7 G, on compare seulement le siège incliné à 20° et la configuration utilisant le repose tête a 45°, pour des désignations à 45 et 90°. Il existe une très grande différence à ce niveau avec seulement 39 % des tâches effectuées correctement avec le siège à 20° contre 78% à 45°.

3.3. - Caractéristiques du déplacement de la tête sous facteur de charge.

Nous n'aborderons ici que les aspects liés à l'acquisition de cibles présentées avec une excentricité limitée. L'amplitude des mouvements de tête est donc beaucoup moins importante que lors de la précédente étude. Il s'agit en fait de saccades de tête obliques vers une localisation prédictible. Pour l'ensemble des sujets et les différents niveaux de facteur de charge les résultats indiquent assez clairement que, dans le domaine d'accélération étudié, la vitesse de déplacement de la tête est très faiblement influencée par le facteur de charge lorsque celui-ci est établi à un niveau donné. Lorsque les acquisitions sont effectuées au début d'une variation d'accélération, les vitesses de déplacement demeurent pratiquement identiques, mais l'on observe une moins bonne stabilité de la visée à l'issue de la saccade.

Si l'on considère maintenant les valeurs moyennes des vitesses crête de déplacement de la tête (fig.5), on constate qu'il n'existe pratiquement pas de différence entre les différents niveaux d'accélération. Par contre la vitesse en gisement est généralement supérieure à la vitesse en site.

L'examen des figures 6 et 7 permet d'affiner le jugement sur les vitesses de saccades de tête. Les tracés ont été obtenus avec une cible se déplaçant dans le plan vertical de haut en bas. On observe que l'acquisition de référence à 1 G (fig.6) met en jeu une seule grande saccade. Par contre à 3 G, (fig. 7) on constate l'existence d'une première saccade, dont la vitesse est identique à la précédente, qui ne permet pas d'atteindre la cible. Une deuxième saccade de correction amène alors la visée sur le but.

Bien que l'intervention de phénomènes adaptatifs survienne rapidement, l'acquisition de la cible par plusieurs saccades successives a été couramment relevée lors des essais.

4. - DISCUSSION

L'ensemble des résultats obtenus lors de ces études successives amène à considérer plusieurs points.

Il apparaît en premier lieu que les caractéristiques de déplacement de la tête ne sont réellement affectées que pour des excentricités de cible importantes, du moins dans un domaine d'accélérations modérées jusqu'à 5 G. Si l'on s'intéresse aux résultats obtenus lors des essais les plus récents, qui analysent plus finement le déplacement de la tête en utilisant des variables cinématiques, on retrouve des observations cohérentes avec la mesure des temps d'acquisition. Dans les deux cas les différences entre les différents niveaux de facteur de charge demeurent très faibles.

Sur le plan du contrôle de la motricité de la tête, on constate donc que les accélérations stabilisées en plateau n'affectent que très modérément les caractéristiques des mouvements de tête, ceci dans un domaine d'accélération et d'excentricité de cible relativement réaliste pour l'emploi d'un viseur de casque.

On peut donc considérer que, dans ce domaine, toutes choses étant égales par ailleurs, le système ostéo-musculaire et articulaire du cou compense presque totalement les effets liés à l'augmentation d'intensité du vecteur gravité. Ces constatations sont à rapprocher des résultats obtenus antérieurement, en particulier par SHIRACHI et coll. (7) qui ont montré que les caractéristiques spectrales dynamiques du mouvement de la tête n'étaient pas affectées par le poids de l'équipement de tête (de 900 g à 2000 g).

En revanche, les variations d'accélérations + Gz semblent poser un problème plus sérieux au système de contrôle moteur de la tête. Les données recueillies lors des différentes campagnes d'essai, ainsi que d'autres provenant d'études complémentaires non présentées ici, convergent sur ce point avec des observations beaucoup plus fondamentales effectuées par VIVIANI et BERTHOZ (10). En utilisant des forces de faible amplitude appliquées dans le plan saggital de la tête avec un domaine de fréquence très large, jusqu'à 20 Hz, ces auteurs ont montré que la réponse biomécanique de la tête, lorsque le sujet résiste activement à la force, était très différente selon les fréquences. Alors qu'en basse fréquence il développe une force opposée à la perturbation, lorsque la fréquence augmente la résistance fait plutôt appel à un raidissement du système musculaire du cou.

Ces expérimentations ont, entre autre, bien montré qu'une analyse statique portant sur les forces et couples maximaux développés par la tête n'étant pas suffisante pour rendre compte des phénomènes dynamiques.

De plus, les observations effectuées à propos de cette expérimentation ont également mis en évidence que la préparation du sujet à subir la stimulation avait une forte influence sur la réponse de la tête. Dans le même ordre d'idée, GUITTON a démontré que certains mécanismes réflexes impliqués dans le contrôle moteur de la stabilité de la tête étaient influencés par le contrôle volontaire (3).

Ceci permet d'aborder indirectement le problème de la traumatologie cervicale liée aux accélérations + Gz. Les observations cliniques, tant en vol, que dans des situations comme le choc à l'ouverture des parachutes, montrent que les problèmes de traumatologie cervicale apparaissent préférentiellement lorsque la "situation subie" diffère de la "situation attendue".

L'ensemble de ces faits semble donc impliquer que l'influence d'un certain niveau de contrôle volontaire est nécessaire pour la mise au jeu optimale des mécanismes visant à stabiliser la position de la tête. Ceci amène à poser le problème des lois de pilotage de l'appareil, surtout pour les avions munis de commandes de vol électriques. On peut se demander si un bon nombre de problèmes rencontrés sur certains avions modernes ne sont pas liés au fait que ces lois n'ont pas pris en compte les provisions nécessaires à la mise en jeu correcte des mécanismes de protection naturels.

CONCLUSIONS

Les résultats obtenus lors de ces différentes études montrent que l'exécution de mouvement de tête de grande amplitude sous fort facteur de charge est notablement améliorée par un siège incliné lorsque l'on utilise l'appui-tête.

Dans un domaine d'emploi plus réaliste des viseurs de casque, on constate que le contrôle de la motricité du cou et de la tête compense correctement les effets du facteur de charge stabilisé si l'on se réfère aux vitesses maximales des saccades d'acquisition.

Enfin, les variations d'accélération sont susceptibles de perturber la stabilité de la tête d'une manière très significative. Les mécanismes impliqués dans la stabilisation de la tête étant notablement influencés par le contrôle volontaire, les lois de pilotage des appareils munis de commande de vol électriques pourraient constituer un point intéressant à considérer.

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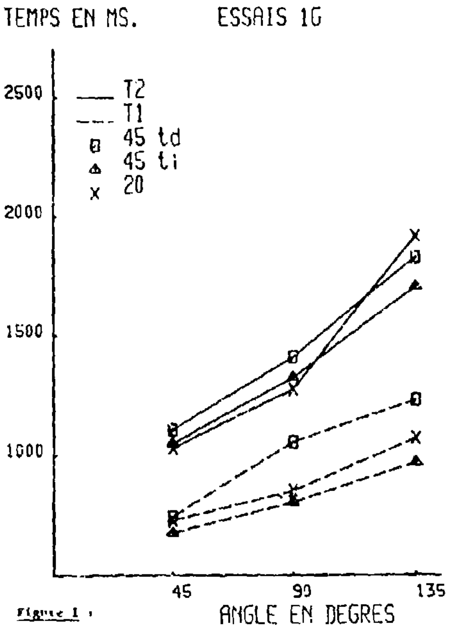

Figure 1 :
Temps d'acquisition de cibles en fonction de l'excentricité et de la configuration du siège.

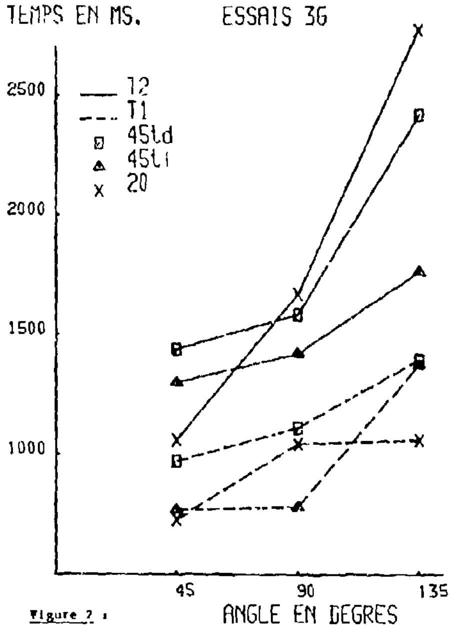

Figure 2 :
Temps d'acquisition de cibles en fonction de l'excentricité et de la configuration du siège.

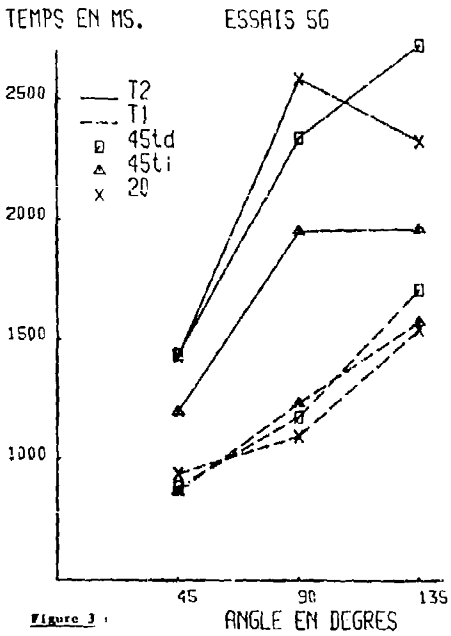

Figure 3 :
Temps d'acquisition de cibles en fonction de l'excentricité et de la configuration du siège.

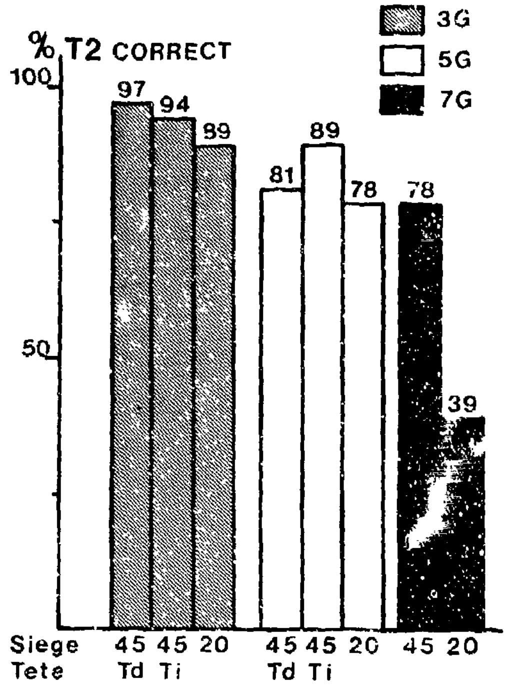

Figure 4 : Pourcentage d alignements effectués correction en fonction du facteur de charge et de l'inclinaison du siège.

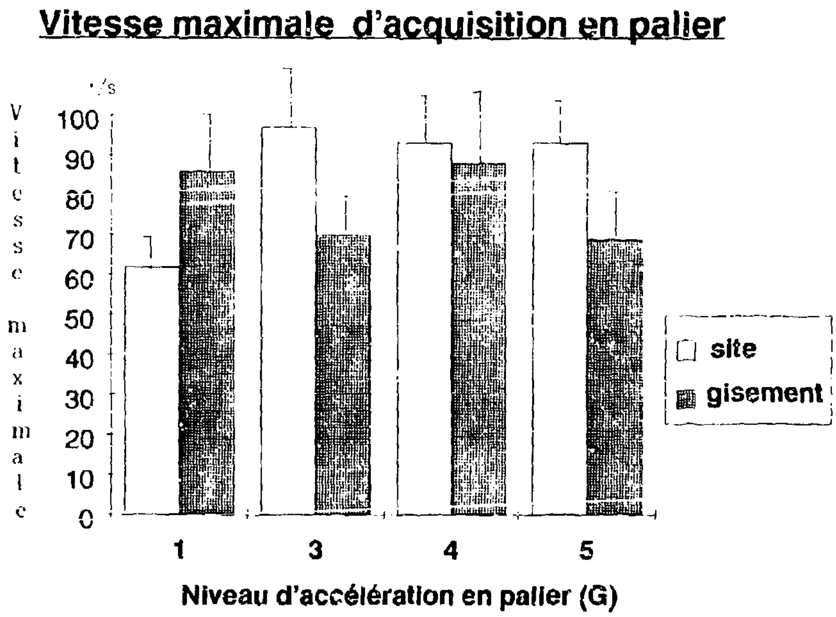

Figure 5 : Valeurs moyennes de la vitesse maximale d'acquisition d'une cible située à + 30° en site et en gisement en fonction du niveau d'accélération.

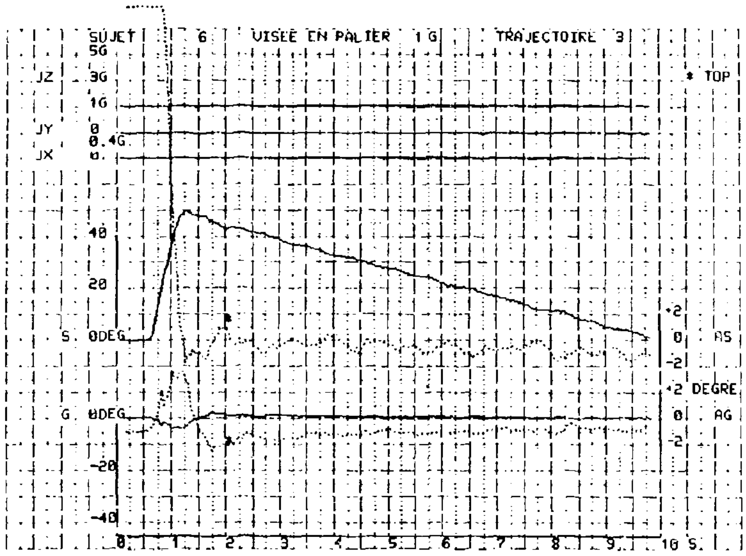

Figure 6 : Acquisition d'une cible se déplaçant dans le plan vertical : Mouvement de la tête effectué en condition de référence à 1 G.

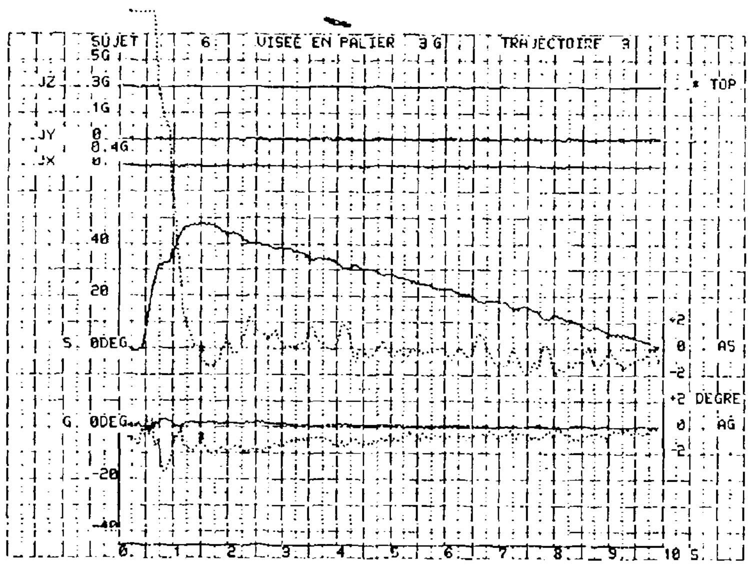

Figure 7 : Acquisition d'une cible se déplaçant dans le plan vertical : sous 3 G, la vitesse initiale de la saccade n'est pas modifiée mais son amplitude est insuffisante.

NECK INJURY PREVENTION POSSIBILITIES IN A HIGH-G-ENVIRONMENT EXPERIENCES WITH HIGH SUSTAINED $+G_z$ TRAINING OF PILOTS IN THE GAF IAM HUMAN CENTRIFUGE

by

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SUMMARY

New generations of high performance military aircraft are able to produce higher G-rates of onset, attain higher G-levels for prolonged periods and in the future will confront man and machine more often with changing acceleration peaks than previous fighter generations have. These enhanced performance capabilities will require additional anti-G-protection equipment as well as special education and training of pilots to tolerate high-G-environment levels so they can fulfill complex tasks during special inflight conditions without suffering from G-induced cardiovascular, pulmonary, cerebral or muskuloskeletal problems.

Therefore an extensive study in 238 young G-stress-unexperienced pilot candidates (aged between 18 - 24 years) of the GAF-Officers'-Academy was performed within a period of 23 months (1 Oct 1985 - 1 Sep 1987) who participated in a high-G training program as volunteers with more than 1250 centrifuge rides partially up to +8 G_z for 30 sec on the GAF IAM Human Centrifuge. By means of an anonymous questionnaire answered by the pilot candidates immediately after G_z exposure and post-acceleration check-up, different data were obtained.
The intension of this investigation was maintaining vision and consciousness as well as neck injury prevention at high sustained $+G_z$ levels in G-unexperienced pilot candidates. The findings in our search for methods to protect a subject under $+G_z$ stress in the human centrifuge from G-induced symptoms, especially of potential cervical spine problems, are described.

INTRODUCTION

Meanwhile there are obviously many worldwide activities in the development of new life support systems to reduce the pilot's straining efforts required to maint..n consciousness at high $+G_z$ stress levels. High-G air combat manoeuvers with frequent tilting and turning of the pilot's head and neck lead to a significantly increasing stress on his cervial spine, especially with the additional weight of the helmet which is further increased by helmet-mounted displays etc.
Even if the vertebral spine is that portion of the musculoskeletal system which sustains the most severe stress, it is the cervical spine which is more susceptible to $+G_z$ stress-induced lesions or injuries, since in the cockpit-environment the cervical column suffers much greater deviations from the vertical alignment than any other part of the vertebral column. It is also well known that column strength and stability is reduced when flexion or torsion movements set in.
Therefore the flight surgeon must contact fighter pilots especially after basic fighter manoeuvers so he can register complaints or injury syndromes like muscle pain or tenseness with painful radiation, muscle spasms and motor or sensory deficits.

MATERIAL AND METHODS

It is also very portant to evaluate these symptoms in G-experienced pilots or G-unexperienced pilot candidates during $+G_z$ stress exposure in the human centrifuge.

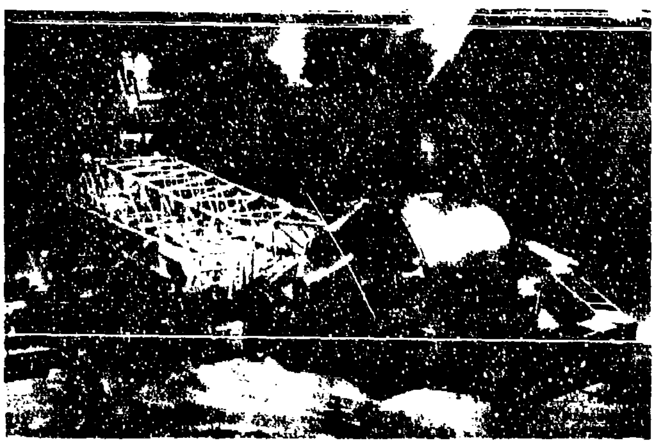

Figure 1: View of the Human Centrifuge at the GAF Institute of Aerospace Medicine in Fuerstenfeldbruck

For that reason the centrifuge rider gets an additional check-up in the post-acceleration period performed by medical personnel to find out if G-induced symptoms occured, i.e. petechias, exhaustion, sweat, motion-sickness-symptoms or pain of the vertebral column, tenseness, muscle spasms or problems in other organs.
In spite of the fact that many pilots fail to seek medical attention for these symptoms-unless the injuries are incapacitating their flying capabilities - and further considering that only a few cases of cervical spine injury resulting from high $+G_z$ exposure have been documented like ligamentous tears, nucleus pulposus hernias or even vertebral fractures, there are potential risks of acute injury and possibly longterm degenerative effects. Beside the observation of the pilot by the flight surgeon during high G-training in the GAF IAM Human Centrifuge monitoring biomedical data like EKG-wave, heart frequency and other vegetative parameters and beside the training of the pilots by the instructor to perform excellent anti-G-straining manoeuvers (AGSM), he is under permanent observation and - if necessary - the pilot's attitude and head position under high sustained G's are corrected to prevent the previously mentioned neck symptoms or cervical injuries.

The data for this presentation were obtained in an extensive study in 235 young G-stress-unexperienced pilot candidates (aged between 18 - 24 years) of the GAF-Officers'-Academy within a period of 23 months (1 Oct 1995 - 1 Sep 1997) who participated in a high-G training program as volunteers with more than 1050 centrifuge rides partially up to + 8 G_z for 30 sec on the GAF IAM Human Centrifuge. The reclining position of the seat of the centrifuge gondola is only 17° which means that G-stress acts exclusively on the pilot's z-axis (vertical axis) and is not partially redirected into his horizontal axis (x-axis) like on a high seat back angle of 30° as realized in the F-16 military aircraft.

By means of an anonymous questionnaire answered by the pilot candidates immediately after G_z exposure and post-acceleration check-up, different data were obtained.

They concern:

- their general feeling after $+G_z$ exposure
- the effects of their AGSM with respect to G-induced visual symptoms
- possible symptoms of motion sickness (especially during the starting- and stopping phase of the circular movements of the centrifuge
- sudden stops of the centrifuge gondola
- physical problems of different organs including the vertebral column, and
- especially the cervical spine.

At 2-hour intervals a group of 3 - 4 pilot candidates was briefed how to handle the centrifuge gondola, how to perform AGSM efficiently, about basics of human physiology and changes under $+G_z$ stress and about management of unexpected emergency situations. Then everybody got 3 rides on the human centrifuge according to the following G-profiles: + 3 G_z for 15 sec to become familiarized with the centrifuge and the effects of G-forces. This was followed by an evaluation of the individual's natural G-tolerance value (his visual endpoint) first without and then during the performance of AGSM. By getting the difference between both values the amount of the individual's personal increase in G-tolerance by AGSM was determined. Then every pilot candidate was subjected to 3 peak profiles consisting of + 3 G_z for 15 sec, + 2 G_z for 15 sec, + 5 G_z for 30 sec - followed by + 2 G_z for 30 sec - and finally + 7 G_z for 15 sec.

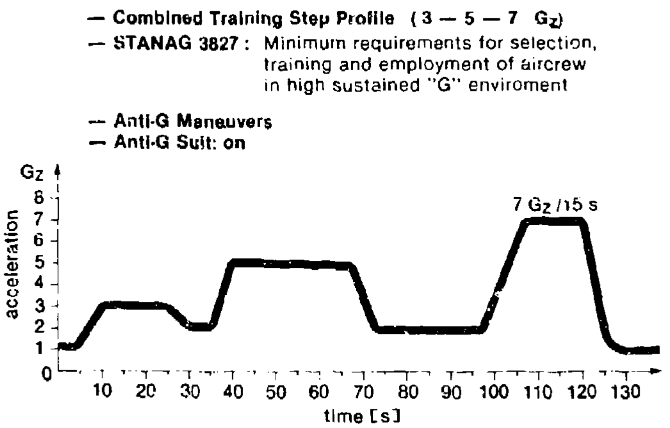

Figure 2:

All these G-profiles were applied by the computer in passive mode for the trainees. The last profile was applied in active mode. This means they followed the G-profile presented on a scope in front of them which consisted of 2 peaks with + 5 G_z by moving the stick. Those pilot candidates who missed the first chance were given a second one. Those who did the centrifuge runs during the first test period successfully were also centrifuged a second time to find out if they could tolerate + 8 G_z for 30 sec.
Every trainee was dressed with his personal flying suit and with an anti-G-suit consisting of 5 bladders. They were pressurized automatically at + 2,2 G_z already. There was voice contact via ear phones from the control room to the centrifuge rider, who did not wear a flying helmet. It was necessary that he always maintained an upright position on the 17° back angle seat and did not perform any head or body movements during $+G_z$ stress to prevent injuries of his cervical spine.

RESULTS AND DISCUSSION

Of a total of 238 pilot candidates 166 pilot candidates or 69.75 % were successful in G-training and fulfilled the minimum requirements of the NATO standardisation agreement, the STANAG 3827 (Minimum Requirements for Selection, Training and Employment of Aircrew in High Sustained "G" Environment (effective as of July 7, 1991)) by tolerating + 7 G_z for 15 sec already in the first test period. This means that 72 pilot candidates (30.25 %) failed. Out of this group which missed the aim (30.25 % of all tested) there were 2 pilot candidates (0.84 % of the total) who took a second chance voluntarily and met the STANAG 3827 requirement.

This means that a total of 168 pilot candidates or 70.59 % of the total number finally were successful. There were also 35 pilot candidates (14.71 % of the total) or 20.83 % who belonged to the successful group (70.59 %) who took a second try on the centrifuge and tolerated + 8 G_z for 30 sec.

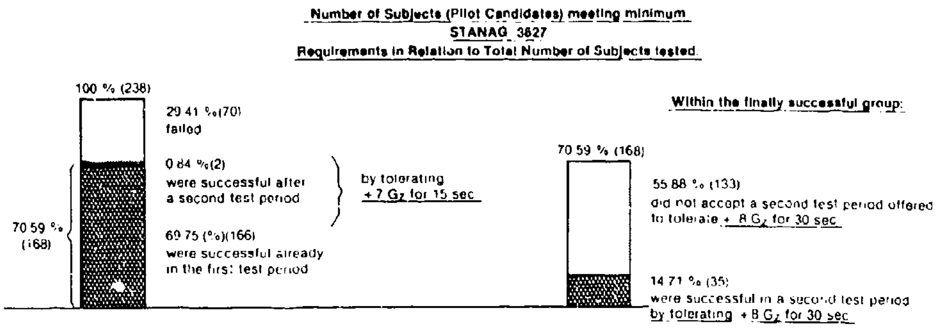

Figure 3:

133 pilot candidates (55.88 % of the total) or 79.17 % who belonged to the group which tolerated + 7 G_z for 15 sec did not accept a second test period offered to tolerate + 8 G_z for 30 sec.

The average G tolerance increase by performing AGSM was + 1.6 G_z in all pilot candidates.
Reviewing the data from the questionnaire and post-acceleration check-up to the 238 pilot candidates in the study there are several interesting findings.

The most impressive event for the pilot candidates in the centrifuge was the G-stress itself. Everybody found the influence of acceleration to be stressful. About 70 % of the candidates had the feeling they could tolerate it, but 30 % found it to be extremely tough. During the centrifuge rides 131 pilot candidates or 55 % of the total had no problems. 107 pilot candidates or 45 % of all had some problems. There were 71 subjects (30 % of the total) with various complaints such as sweat, exhaustion etc. and as many as 15 % suffered from motion sickness symptoms like vertigo and nausea and within this group 4.5 % even from vomiting.

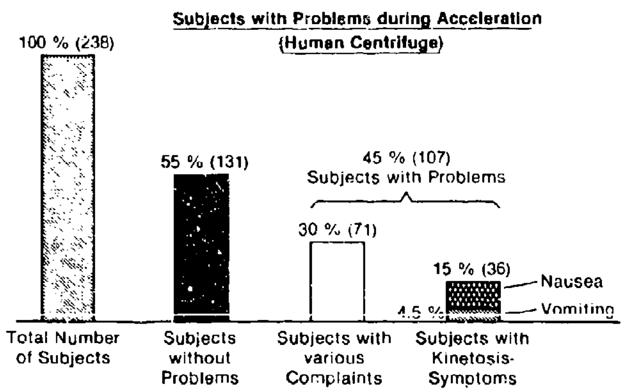

Figure 4:

We found that motion-induced vegetative symptoms such as vestibular illusions made a big impression on the trainees.
Especially in the deceleration phase 214 pilot candidates or 90 % of all had vegetative symptoms. We found major symptoms in 32.5 % and minor symptoms in 57.5 % compared to the acceleration phase where 10 % had vestibular illusions but of less severity. 9.6 % had moderate and only 0.4 % had strong symptoms.

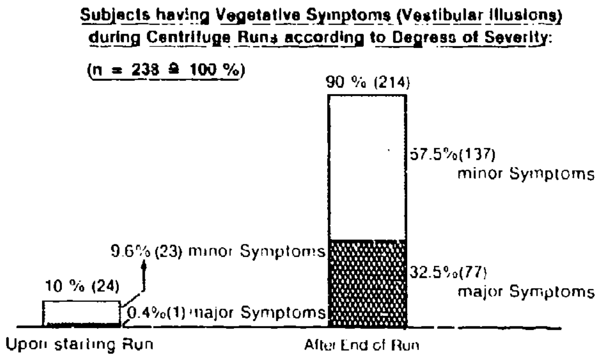

Figure 5:

The data of the questionnaire also revealed that within this group of 238 G-unexperienced pilot candidates 89.5 % (213 subjects) had a tunnel vision or greyout, 9.0 % (21 subjects) had a shortlistening blackout and 1.5 % (4 subjects) had a G-LOC.

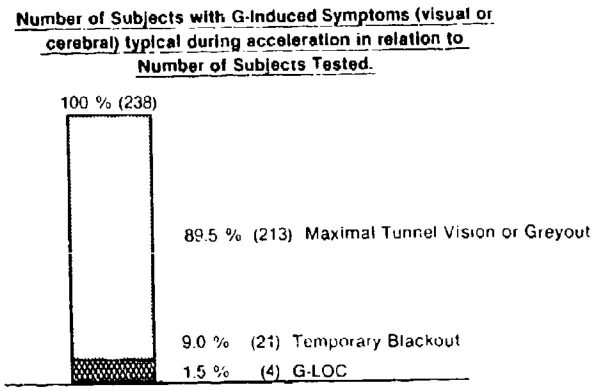

Figure 6:

It was also interesting to note that 10 % of the whole group (24 subjects) complained of pain, muscle spasms or tenseness in the area of the lower and upper legs, in the muscles of the abdominal and the cervical area. 9 pilot candidates or 3.6 % of the whole group (238 subjects) which is according to about one third of this affected group (24 subjects) reported neck affection.

Number of Subjects with musculo-skeletal Problems
After Acceleration in the Human Centrifuge:

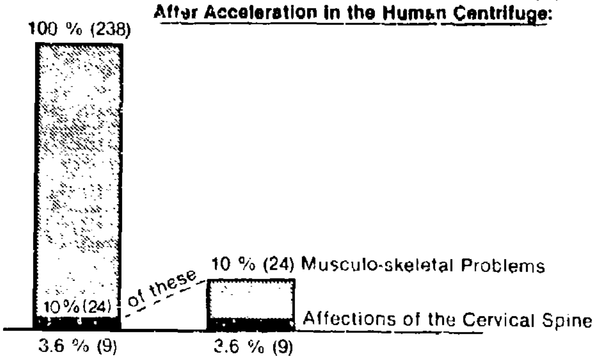

Figure 7:

Even though there was no case of cervical spine injury it is remarkable that minor cervical problems did occur under G-stress up to + 9 G_z, despite the fact that the trainees were in an upright position, did not wear helmets and did not move their heads under G-loads.

CONCLUSIONS

1. G-unexperienced subjects such as pilot candidates should have their first experience with G_z-stress without wearing a flying helmet, in an upright position and without moving their heads
2. They should wear their flying helmet and move their head only after some initial experience with "G's". Therefore we recommend a simulation of air combat manoeuvers for pilot candidates under G-loads at a later stage of G-training in the centrifuge to prevent cervical spine injuries. Only combat-ready pilots participating in a high-G-training program in the human centrifuge should wear their helmets and should perform routine head movements like the "check-six" manoeuver by looking back over their shoulder under G-loads
3. A seat having a small back angle must be recommended even if the potential benefit of an increased reclination of 30° as in the F-16 aircraft is lost, because it is a fact that many G-experienced pilots lean forward during high-G manoeuvering to enhance their search or maintain sight of the attacker in air combat situations. This means that during those periods of hig-G stress their vertebral column will get not support by their 30° reclination seat
3\. In an air combat scenario under high sustained G it is unrealistic for the pilot not to move his head. From time to time he also has to look back over his shoulder "checking-six", a most stressful situation for the cervical spine under G-forces. This calls for an improvement of the support system for the vertebral column, especially for the cervical equipment like helmet-mounted displays etc. Better cervical support systems should be developed in the future.
5. Prior to undergoing high-G-training in the centrifuge or pulling high G-loads in flight pilots should go through a "G-warm-up" of their cervical spine by stretching the neck muscles and they should cautiously control their neck movements under high-G-stress. Participation in a regular neck exercise program to strengthen these muscles can be of great benefit and prevent injuries
6. Finally the high G-fighter pilot should reassume high G-missions only gradually after prolonged interruptions.

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Risque de lésions cervicales en accidents reels et simules

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RESUME

Le Laboratoire de Physiologie et de Biomécanique associé à PSA et RENAULT dispose de deux sources de données lui permettant d'apporter une contribution à la compréhension du risque mais aussi du mécanisme des lésions cervicales. Ces données proviennent :

- de son enquête pluridisciplinaire concernant les occupants de voitures impliqués en accidents corporels qui comporte actuellement 6589 véhicules et 9789 impliqués aux places avant.

- de 375 tests expérimentaux avec cadavres : collisions simulées frontales et latérales, chutes libres, essais avec impacteurs etc...

La première partie sera plus spécialement axée sur le risque de lésions cervicales avec ou sans impact direct de la tête pour les seuls occupants ceinturés impliqués dans différents types d'accidents réels.

La seconde partie concernera les essais avec cadavres qui permettent une meilleure compréhension du mécanisme des lésions cervicales grâce, notamment, aux mesures de différents paramètres physiques (maximum de l'angle tête/thorax, vitesse et accélération angulaire, etc...).

1) LES OCCUPANTS CEINTURES IMPLIQUES EN ACCIDENTS REELS

Nous distinguerons dans ce qui suit deux catégories de lésions cervicales :

- les lésions mineures (AIS = 1)* : douleur, contusion, entorse sans aucune atteinte neurologique,

- Les lesions graves (AIS ≥ 2)* : Fractures des vertèbres cervicales avec ou sans atteinte neurologique,

Parmi les 3781 occupants ceinturés des places avant de notre enquete, toutes configurations d'accidents confondues, on observe seulement 41 cas de lésions graves (soit 1,1 %) et 400 lésions mineures (10,6 %). Les lésions très graves (paraplégie) sont extrêmement rares, 3 cas dont un mortel sont observés dans des chocs divers ou la tête est toujours sévèrement lesée par impact direct.

Pour une meilleure compréhension du risque cervical, nous nous interessons dans ce qui suit aux occupants ayant subi un seul choc et distinguons trois catégories de trajectoires pour les occupants : frontale, latérale et arrière. En l'absence d'impact de la tête, les sollicitations de la colonne cervicale seront respectivement en hyperflexion frontale ou latérale et en hyperextension.

L'influence sur le risque cervical d'un impact de la tête, qui ne peut que modifier la sollicitation du cou est analysée ainsi que l'influence de la violence de la collision.

1.1. Part des lésions cervicales selon le type de choc (tableau 1)

Avant toute étude sur les seules lésions du cou, il n'est pas inutile de préciser la part des lésions cervicales par rapport aux autres territoires corporels pour ces impliqués en accidents de voitures. Toutes vitesses confondues, on observe que :

* AIS : Abreviated Injury Scale (voir annexe 1)

- en choc frontal (60 % des accidents corporels), la fréquence de lésions du cou est au 5ème rang (10,2 %) après les membres inférieurs (34 %), le thorax (31,5 %), la tête (31,2 %) et les membres supérieurs (20,7 %).

- en choc latéral (20 % des accidents corporels), 8,4 % des occupants présentent des lésions du cou, soit l'avant dernier rang.

- en choc arrière (10 % des accidents) le cou figure au 1er rang avec 26,1 % de lésions mineures et 1,31 % de lésions cou plus sévères. Viennent ensuite la tête (17,0 %) et la colonne dorsolombaire (13,7%).

En ce qui concerne les fractures, la fréquence est de l'ordre de 1 % quel que soit le type de choc et pour ce niveau de sévérité de blessures (AIS $\geq$ 2), le cou se place pour les chocs frontaux et latéraux, respectivement aux 7è et 8è rang avant et après la colonne dorsolombaire. En choc arrière, où la gravité globale de l'occupant (nombre de lésions par occupant) est plus faible que dans les autres chocs, la tête puis le cou sont prioritaires, mais il est vrai que le choc arrière ne représente seulement que 4 % des blessés graves et tués.

1.2. Fréquence des lésions cervicales selon le type de choc

1.2.1. Le choc frontal

Avant toute analyse sur les lésions cervicales, il est utile de préciser que sans impact de la tête, on observe seulement 2 cas de brèves pertes de connaissance parmi les 1112 impliqués, il n'existe donc aucun risque pour la tête en absence d'impact et ce, quelle que soit la violence de la collision exprimée par la variation de vitesse subie par l'occupant au cours du choc (ΔV).

S'agissant du cou, la retenue thoracique par la ceinture (qui cree parfois des fractures de côtes pour les plus âgés) entraîne sans impact de la tête, un mouvement d'hyperflexion du cou qui peut générer des contraintes se traduisant dans 10% des cas par des douleurs d'origine musculo-ligamentaires, comme l'indique le tableau suivant. Ce n'est que dans 0,5 % des cas qu'une lésion plus sévère peut être observée (antécedents ').

En cas d'impact de la tête, la part des lésions graves est plus importante mais reste encore très faible, inférieure à 3 %.

Tableau 1	Pas d'impact de la tête AIS tête = 0 N = 1112	Impact de la tete	
		AIS tête = 1 N = 311	AIS tête $\geq$ 2 N = 193
Lésions du cou AIS cou $\geq$ 1	108 9,7 %	32 10,2 %	24 12,4 %
Lésions graves du cou AIS cou $\geq$ 2	5 0,45 %	6 1,9 %	5 2,6 %

Selon les niveaux d'AIS à la tête, les occupants correspondants ne sont pas impliqués dans des chocs de même violence (Figure 1). Ce biais de violence pourrait alors expliquer à lui seul le risque supplementaire de fractures de vertebres cervicales pour les cas avec impact de la tête. Une étude plus precise par classes de vitesse fait apparaître (tableau 2) :

* Les fréquences de lésions du cou (AIS $\geq$ 1) avec ou sans impact de la tête n'augmentent pas significativement* avec la violence du choc contrairement aux fréquences des lésions tête (en cas d'impact) qui augmentent fortement avec le ΔV (fig.2).

* Dans une même classe de vitesse, un impact de la tete n'augmente pas significativement la fréquence de lésions du cou (AIS $\geq$ 1) mais augmente le risque de fractures (AIS $\geq$ 2) pour les chocs supérieurs à 25 km/h.

L'absence de corrélation entre risque pour le cou et violence de choc pour chacun des 2 groupes avec et sans impact de la tête (fig.3) est due à divers facteurs tels que :

- différence de forme et de rigidité des éléments impactés par la tête (volant, planche de bord) ;
- impacts plus ou moins tangentiels de la tête contre ces éléments ;
- tolérances interindividuelles très différentes ;
- influence de l'âge et du sexe ;

Certains de ces facteurs, impossibles à quantifier en accidents réels pourront grâce aux essais avec cadavres être analysés avec beaucoup plus de précision ;

* Test du X2 significatif au seuil de 0.05

1.2.2. Le choc lateral

Compte-tenu du nombre très faible de lesions graves du cou (3 cas), il n'est pas possible de conclure quant à la gravité observée avec (2 cas) ou sans impact de la tête (1 cas) ; on peut tout simplement noter que ces fractures de la colonne cervicale sont observées au-dessus de 30 km/h de ΔV et que, la fréquence d'apparition, des lésions est plus importante avec que sans impact de la tête (différence toutefois non significative).

	Pas d'impact tête	Impact tête	
Tableau 2	AIS tête = 0 N = 225	sans fracture sans perte de connaissance AIS tete = 1 N = 76	avec fracture et/ou perte de connaissance AIS tete $\geq$ 2 N = 44
Nombre de lésions du cou AIS cou $\geq$ 1	16 6,7 %	8 11,7 %	5 11,4%
Nombre de lésions graves AIS cou $\geq$ 2	1 0,44%	1 1,3 %	1 2,3%

L'analyse par classes de vitesse donne des résultats de meme ordre que pour les occupants impliqués en choc frontal, à savoir :

- le risque pour la tête est fortement lie à la violence du choc alors que celui du cou (AIS $\geq$ 1) ne varie pas qu'il y ait impact de la tete ou non (Figures 4 et 5).

- pour une meme classe de vitesse il n'y a pas de difference significative de risque pour le cou avec ou sans impact de la tête.

En choc latéral, selon la place de l'occupant par rapport au point de choc, deux configurations sont possibles :

- occupants places du cote opposé au choc : l'ensemble tete-cou-thorax est libre dans son mouvement latéral, pas d'arrêt par la ceinture au niveau de la colonne cervicale.

- occupants placés du côté du choc : dans la quasi totalité des cas l'occupant bénéficie d'un appui de l'épaule et du thorax contre sa porte adjacente qui limite ainsi un deplacement latéral qui, important pourrait être prejudiciable a la colonne cervicale. De plus la vitre latérale ou le pied-milieu (B pillar) peut réduire l'amplitude du mouvement de rotation de la tête. Si l'ensemble tête-cou-thorax n'est arrête que par la ceinture de sécurite au niveau du cou (par exemple : ouverture de la porte adjacente) les lesions peuvent être severes : c'est le cas de la seule lésion grave du cou observee sans impact de la tête. Une implantation inversee des ancrages actuels (sans mesure adjacente de protection) aurait certainement des conséquences gravissimes pour les impliqués en choc latéral (1). Des essais avec cadavres sont en cours pour verifier l'influence de cette position d'ancrage sur le risque pour le cou.

1.2.3. Le choc arrière

En choc arrière unique, 2 cas seulement (1,3 %) de fractures cervicales sont observés. La distribution des risques pour le cou (AIS $\geq$ 1) pour les 153 impliqués est indiquée dans le tableau 3.

	AIS cou $\geq$ 1	
	< 25 km/h	> 25 km/h
Tête AIS = 0	25/93 26,9 %	11/34 32,4 %
Tête AIS $\geq$ 1	2/12 16,7 %	2/14 14,3 %

(1) Référence en annexe.

Beaucoup de publications ont eu pour thème les lésions du cou en choc arrière, nous nous limiterons donc aux conclusions essentielles car beaucoup de facteurs interviennent en choc arrière tels que la présence d'un appui-tête, la tenue du dossier de siège, la violence du choc, le sexe de l'occupant, la distance tete-appui-tête...

- La détérioration du siège apparaît plus performante que l'appui-tête pour la réduction des douleurs cervicales.
- Cette détérioration observée dans des chocs généralement plus sévères, qui autorise des heurts mineurs de la tête permet par contre de réduire la fréquence du risque de douleur cervicale (tableau précédent).
- L'efficacité de l'appui-tête est nulle pour les hommes et très faible pour les femmes dont la fréquence des douleurs cervicales est proche du triple de celle des hommes.

Dans l'hypothèse d'un siège ne se détériorant pas, l'appui-tete est alors indispensable et la distance tête/appui-tête est alors prépondérante. Cette distance qui sur la plupart des véhicules actuels est importante explique certainement l'efficacité quasi nulle de l'appui-tête quand le siège est intact. Des essais sont actuellement en cours avec cadavres et mannequins pour tenter de comprendre les sollicitations du cou en faisant varier la distance tête/appui-tête.

1.3. Typologie et positionnement des lésions cervicales graves

Deux des 21 occupants blessés gravement à la colonne cervicale ne présentent pas de fracture mais des signes de souffrance neurologique (paresthésie). Plus de la moitié (11 sur 19) des occupants présentant des fractures sont lésés au niveau de C2, les autres fractures se répartissant entre C3 et C7 (figure 6).

Sans impact de la tête, cinq des 7 fractures se situent au niveau de C2.

1.4. Influence du sexe sur la fréquence des lésions cervicales

Le sexe de l'occupant semble déterminer pour beaucoup la probabilité d'une atteinte cervicale. Ainsi dans les trois configurations de choc décrites précédemment, la fréquence des lésions cervicales (AIS $\geq$ 1) chez la femme est plus du double de celle de l'homme qu'il y ait impact ou non de la tête.

La différence de morphologie cervicale et de positionnement relatif de l'appui thoracique pourraient expliquer cette plus grande vulnérabilité de la femme.

Type de choc	Fréquence des lesions du cou en % AIS $\geq$ 1	
	Femmes	Hommes
Frontal	16,9	7,5
Latéral	13,0	6,5
Arriere	52,0	18,4

Tableau 4

S'y ajoute qu'en choc arrière, la moindre masse de la femme (en moyenne) ne lui permet pas de déformer le dossier a des vitesses pour lesquelles ces déformations se produisent pour les hommes.

2) ESSAIS AVEC CADAVRES

De nombreux essais ont été réalisés avec sujets humains à partir de 1973 car, il était apparu rapidement que l'évaluation de la qualité de la protection offerte aux accidentés de la route, par l'utilisation des mannequins de choc, n'était pas suffisamment fiable, en raison de leur médiocre simulation du comportement humain et des lacunes des connaissances sur la tolérance humaine aux chocs, notamment pour le cou.

Le Laboratoire de Physiologie et de Biomécanique entreprit une coopération étroite et continue avec des instituts médicaux spécialisés dans les expérimentations avec sujets humains pour effectuer des essais susceptibles de fournir les données biomécaniques nécessaires

Une méthodologie très stricte a été mise au point pour pouvoir interpréter avec suffisamment de précision les résultats d'essais effectués avec des cadavres frais ;

- enquête sur les causes du décès,

- mesures anthropométriques très précises,
- caractérisation osseuse des sujets comparativement aux vivants,
- rétablissement de la pression dans le système vasculaire avec un liquide chargé de particules de carbonne qui permet :
  . de restituer le réalisme de la dynamique des organes internes (masses, raideurs, volumes).
  . d'éviter le "découplage" du cerveau par rapport au crâne en rétablissant une pression moyenne au niveau du tissu cérébral.
  . la visualisation des atteintes vasculaires (y compris des ruptures capillaires cérébraux.)
- rétablissement de la pression pulmonaire pour une remise en position normale des viscères intrathoraciques notamment diaphragme et poumons.

Cette méthodologie utilisée pour la plupart des essais permet une meilleure compréhension du mécanisme de différentes lésions : deflexions thoraciques et fractures de cotes, lésions abdominales par "sous-marinage" avec ceinture de sécurité ou par impact direct en choc latéral, lésions crânio-cérébrales, fractures de la colonne vertébrale et notamment fractures de vertèbres cervicales.

En choc frontal et en choc latéral, de nombreux essais ont été effectués soit dans des véhicules complets, soit sur catapulte. Les résultats de ces essais en ce qui concerne le risque de fractures cervicales sont décrits ci-dessous.

2.1. Le choc frontal

Cent cadavres porteurs d'une ceinture 3 points ont été testés en choc frontal à des vitesses comprises entre 50 et 65 km/h et pour des décélérations moyennes de 10 à 25 g (17g à 50 %).

Comparativement, les véhicules impliqués en accidents réels pour une même classe de vitesse, subissent des décélérations plus faibles (environ 13 g).

Cette différence importante de décélération influe-t-elle sur le risque de fractures cervicales ? Le tableau suivant indique que, pour les essais avec cadavres, on n'observe pas de différence significative de risque pour deux niveaux de décélération. Cependant d'autres conditions d'essais telles que : type de retenue (avec ou sans limiteur d'effort, avec ou sans pretension), sexe .... pourraient contribuer à expliquer ce résultat.

	Décélération véhicule		
	10 à 16 g	17 à 25 g	TOTAL
Avec impact tête	6/18 33,33 %	3/19 15,8 %	9/37 24,3 %
Sans impact tête	3/38 7,9 %	2/25 8,0 %	5/63 7,9 %

Le fait qu'il y ait impact de la tete multiplie par 3 le risque de fractures cervicales.

Pour des chocs de violence identique, les occupants impliqués en accidents réels présentent des risques de lésions graves du cou beaucoup plus faibles (1,6 % sans impact de la tête, 2,7 % avec impact).

Ce risque plus important pour les sujets humains peut s'expliquer :

- par une plus grande sévérité des impacts tête en tests du fait de trajectoires tres axiales, ce qui n'est pas le cas dans la majorité des accidents reels (impacts plus tangentiels dus a l'obliquité d'un grand nombre de collisions).
- par une possible moindre tolérance de la colonne cervicale chez les sujets humains dont la moyenne d'âge est de 57 ans contre 30 ans chez les accidentés réels pour ces chocs très sévères.

La moindre tolérance du thorax pour les personnes les plus âgées a été verifiée dans notre enquête (figure 7) grâce à des accidentés porteurs de ceintures équipées d'amortisseurs textiles ; le cou pourrait subir la même évolution.

En ce qui concerne les lésions de la tete par impact, on note pour les essais où le système vasculaire du cerveau a été remis en pression que les lésions cervicales apparaissent avec et sans lésion grave à la tête :

- dans 3 cas sur 8, on observe des fractures de vertebres cervicales associées a des lésions graves à la tête (fracture de la face et/ou lésion interne).

../...

- dans 2 cas sur 10, il existe des fractures cervicales sans lesion grave associée à la tête.

D'autre part, sans impact de la tête, aucune lésion du cerveau ou du tronc cerebral n'est observée.

2.2. Le choc latéral

42 sujets ont été utilisés dans des chocs lateraux divers dont la moitié pour des reconstitutions d'accidents réels et 5 autres pour des duplications d'essais effectués avec volontaires aux U.S.A. (essais EWING ΔV = 22 km/h - γ moyen = 7 g et 13 g).

Aucun de ces 26 cadavres ne présentait des fractures de vertèbres cervicales comme dans la realité des accidents. Il est vrai que contrairement au choc frontal où la difference d'âge des 2 groupes est tres importante, en choc latéral pour les reconstitutions d'accidents l'âge des sujets utilisés était relativement proche de celui des accidentes.

Pour les 16 autres tests, on ne note qu'un seul cas de fracture au niveau de l'apophyse articulaire de C7, mais les conditions de choc étaient d'une sévérité excessive par rapport à celles des accidents réels.

S'agissant des lésions tête, on observe pour 28 des 42 sujets injectes correctement, 7 cas de lésion du cerveau dont 3 sans impact de la tête. Pour ces 3 cas, on observe seulement un mouvement de grande amplitude de la tête.

Certaines lésions (amnésie traumatique importante, coma stade I ou II) chez les accidentés où un impact sévère de la tête est difficilement possible contre la vitre latérale, pourraient peut-être s'expliquer ainsi !

2.3. Les essais de chutes

De par leur simplicité, ces essais permettent une analyse plus precise de l'influence de différents facteurs ; ainsi pour chaque essai on mesure :

- l'angle maximum de flexion de la tête par rapport au thorax ;
- l'accélération angulaire de la tête ;
- la vitesse angulaire de la tête ;
- le HIC (Head Injury Criterion) ;

L'autopsie détaillée de la tête et du cou permet de déterminer l'ensemble des lésions y compris celles du cerveau qui a été remis en pression.

* En ce qui concerne les lésions cervicales, les résultats des deux séries d'essais (chutes frontales et latérales) sont les suivantes :

  . en chutes frontales, les fractures apparaissent pour des angles d'hyperextension tête/thorax supérieurs à 65° associés à des vitesses angulaires supérieures à 45 rad/s mais ces fractures ne sont pas systématiques (figure 8) ;

  . en chutes latérales, il y a lésion cervicale au-dessus de 55° d'angle de flexion et de 50 rad/s de vitesse angulaire (figure 9).

  Ces fractures cervicales sont associees à des fractures et/ou lésions du cerveau dans tous les cas en chutes frontales et dans 2 cas sur 6 en chutes latérales.

* S'agissant des lésions du cerveau, 70 % sont situées au niveau du tronc cérébral. Elles sont associées dans 70 % à des fractures du crâne ou de la face en chutes frontales et seulement dans 20 % en chutes latérales.

  Un HIC de 1000 est considéré comme la valeur maximun tolérable avant apparition de blessures graves à la tête, il est donc intéressant pour l'ensemble de ces essais de comparer ces valeurs de HIC calculés pour les cas avec et sans lésion grave à la tête ;

  - en frontal, les HIC varient de 516 à 2351 (moyenne 1232) pour les cas avec blessures graves et de 692 à 2138 pour les cas de blessures mineures (moyenne 1282) ;
  - en latéral, les HIC varient de 641 à 1600 (moyenne 1201) pour les cas de blessures graves et de 899 à 1665 (moyenne 1229) pour les cas de blessures mineures.

  Compte-tenu de ces résultats, la valeur limite de 1000 ou toute autre valeur semble peu crédible pour prédire l'apparition d'une blessure grave. Le HIC n'est donc pas un critère pertinent à lui seul. Quels sont alors le ou les critères pertinents ?

Certainement une combinaison de plusieurs paramètres : HIC, vitesse angulaire, accélération angulaire. Le Laboratoire et plusieurs autres équipes étudient actuellement l'influence de ces derniers paramètres. Une étude effectuée par le Laboratoire avec des boxeurs volontaires montre que des vitesses angulaires de la tête de 45 rad/s (3 ms) et des accélérations angulaires de 8600 rad/s2 (3 ms) sont acceptées sans aucune altération physique et physiologique. Les valeurs maximales correspondantes atteignent 48 rad/s et 16200 rad/s2 (2,3) (Figure 10).

Cette figure démontre aussi que pour deux sujets humains placés dans un même véhicule, (126.1 et 126.2) seule l'accélération angulaire explique la lésion cérébrale grave du 126.1, les HIC proches ne permettant pas ici encore une distinction entre lésion grave et absence de lésion.

Les différents travaux du Laboratoire ont permis entre autre de mettre au point un cou biofidèle qui équipe actuellement le mannequin EUROSID (4,5,6).

CONCLUSIONS

Quel que soit le type de choc, les fractures graves de la colonne cervicale sont rares dans la réalité routière, elles ne sont présentes que pour 1,1 % des accidentés ceinturés.

Sans impact de la tête, le risque est de 0,4 à 0,5 % pour 2 % environ avec impact.

Que les lésions soient mineures ou graves, les femmes présentent 2 à 3 fois plus de risque de lésions du cou que les hommes.

L'apparition des lésions du cou est relativement indépendante de la vitesse et de la décélération moyenne du véhicule au cours du choc qu'il y ait impact ou non de la tête. Ces décélérations moyennes pour les chocs les plus violents sont de l'ordre de 16 à 18 g (maximum possible à 35 g) pour des variations de vitesse de 50 à 65 km/h.

Les essais avec cadavres permettent de vérifier que, en choc frontal :

- sans impact de la tête, il n'y a pas de lésion du cerveau (comme dans la réalité routière) et que le risque de lésion cervicale grave est toujours très inférieur au risque observé avec impact de la tête ;

- avec impact de la tête, ces fractures cervicales apparaissent pour des angles d'hyperextension tête/thorax de l'ordre de 60° associés à des vitesses angulaires élevées 50 rad/s.

En choc latéral, l'association de ces deux mêmes critères, pour des valeurs voisines d'angle de flexion et de vitesse angulaire, détermine l'apparition de fractures de la colonne cervicale.

CHOC FRONTAL 1616 OCCUPANTS

	Tête	Cou	Thorax	Membres Supérieurs	Colonne Dorso Lombaire	Abdomen	Bassin	Membres Inférieurs
AIS 1	504 31,2	164 10,2	509 31,5	335 20,7	74 4,6	118 7,3	102 6,3	549 34,0
AIS 2	193 11,9	16 1,0	87 5,4	99 6,1	17 1,05	41 2,5	29 1,8	163 10,1

en %

CHOC LATERAL 345 OCCUPANTS

	Tete	Cou	Thorax	Membres Supérieurs	Colonne Dorso Lombaire	Abdomen	Bassin	Membres Inferieurs
AIS 1	120 34,8	29 8,4	92 26,7	79 22,9	21 6,1	36 10,4	31 9,0	75 21,7
AIS 2	44 12,7	3 0,87	24 7,0	13 3,8	4 1,2	15 4,4	14 4,1	10 2,9

CHOC ARRIERE 153 OCCUPANTS

	Tete	Cou	Thorax	Membres Superieurs	Colonne Dorso Lombaire	Abdomen	Bassin	Membres Inferieurs
AIS 1	26 17,0	40 26,1	7 4,6	12 7,8	21 13,7	3 2,0	3 2,0	1 6.5
AIS 2	8 5,2	2 1,31	1 0,6	2 1,3	---	---	---	1 0,6

Tableau 1 : Fréquence des lésions selon le type de choc par territoire corporel pour les occupants ceinturés (choc unique).

LESIONS DU COU AIS ≥ 1

Violence du choc	Pas d'impact de la tête AIS = 0	Impact de la tête AIS ≥ 1
25 km/h	47/549 8,6 %	5/67 7,5 %
26-45 km/h	56/501 11,2 %	34/253 13,4 %
45 km/h	5/62 8,1 %	17/184 9,2 %

LESIONS DU COU AIS ≥ 2

Violence du choc	Pas d'impact de la tête AIS = 0	Impact de la tête AIS ≥ 1
25 km/h	1/549 0,2 %	0/67 0 %
26-45 km/h	3/501 0,6 %	7/253 2,7 %
45 km/h	1/62 1,6 %	4/184 2,2 %

Tableau 2 : Fréquence des lésions du cou selon la violence du choc et la présence ou non d'un impact de la tête en choc frontal.

ANNEXE 1

AIS - ABBREVIATED INJURY SCALE

TETE : AIS = 1 Plaie superficielle, contusion, douleur, de la face ou du crâne sans perte de connaissance.

AIS $\geq$ 2 Fractures deplacées ou non des os de la face ou du crâne associées ou non à des lésions du cerveau : perte de connaissance brève ou importante, coma, hématome sous et extra-dural....

COU : AIS = 1 Douleur, contusion, entorse du rachis cervical sans atteinte neurologique.

AIS $\geq$ 2 Fracture des apophyses, pedicules, ou corps de la vertèbre cervicale avec ou sans atteinte neurologique.

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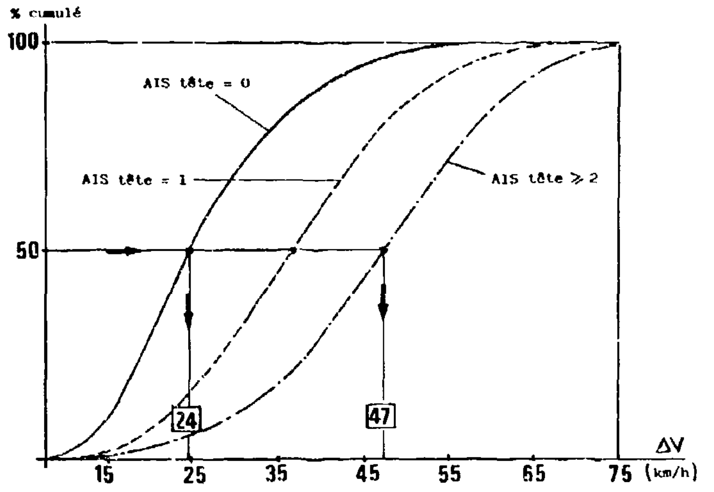


Figure 1. Distribution des violences des chocs en fonction de l'AIS à la tête en choc frontal

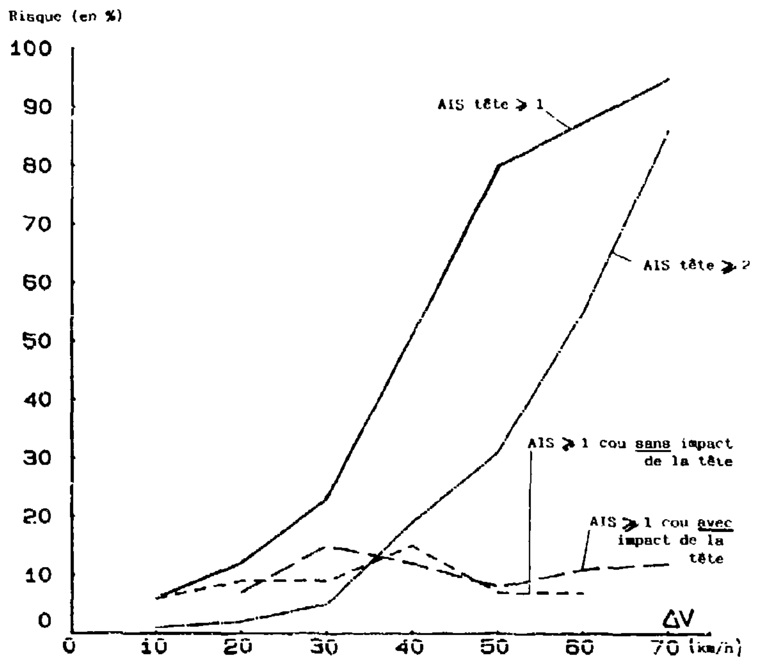
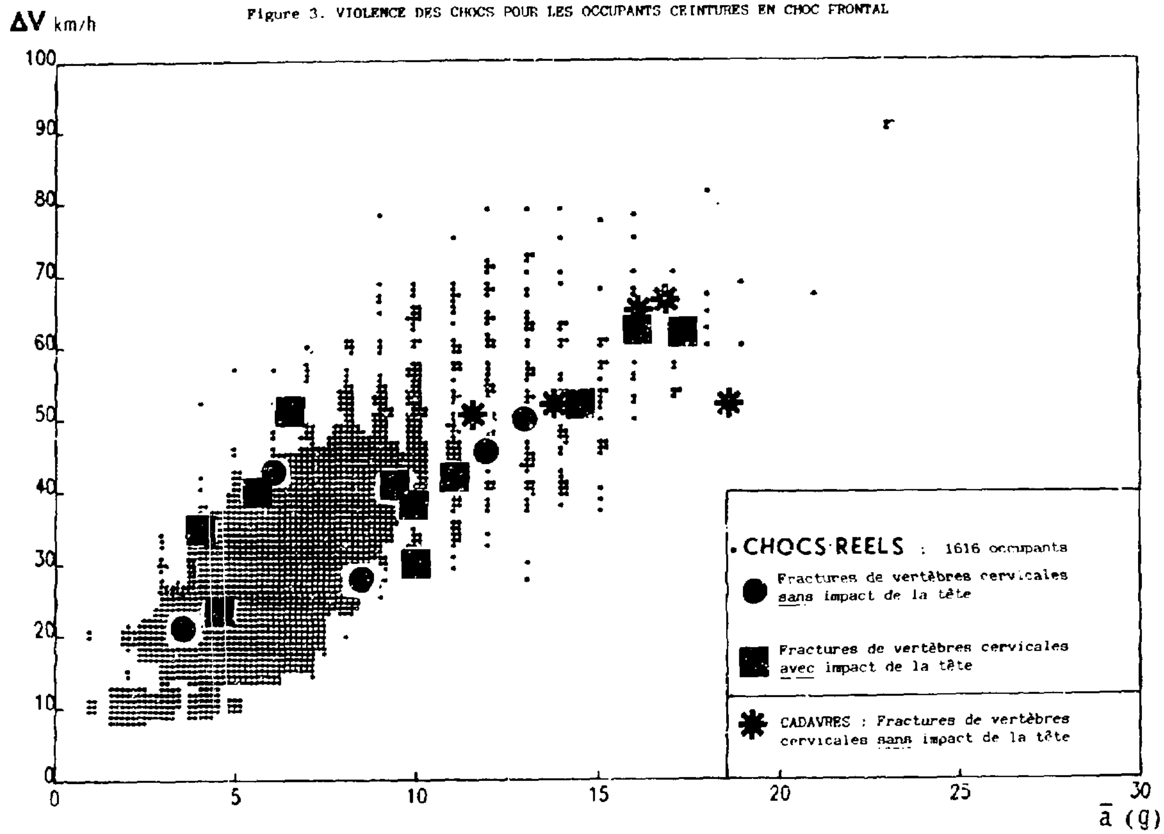


Figure 2. Risque de lésions à la tête et au cou en fonction de la vitesse pour le choc frontal

Figure 3. VIOLENCE DES CHOCS POUR LES OCCUPANTS CEINTURES EN CHOC FRONTAL



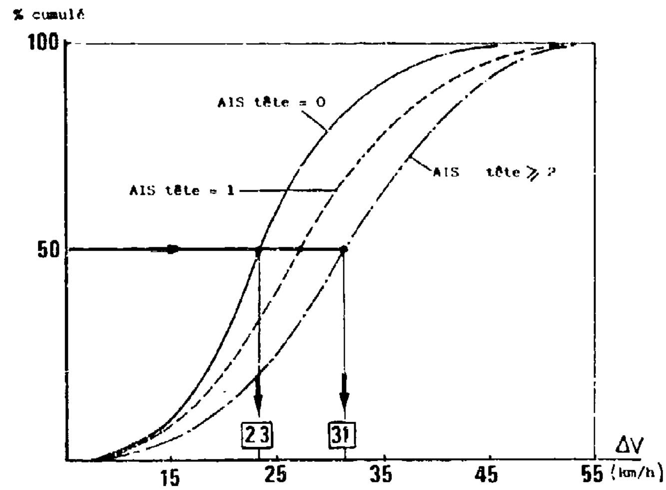


Figure 4. Distribution des violences des chocs en fonction de l'AIS de la tête en choc latéral

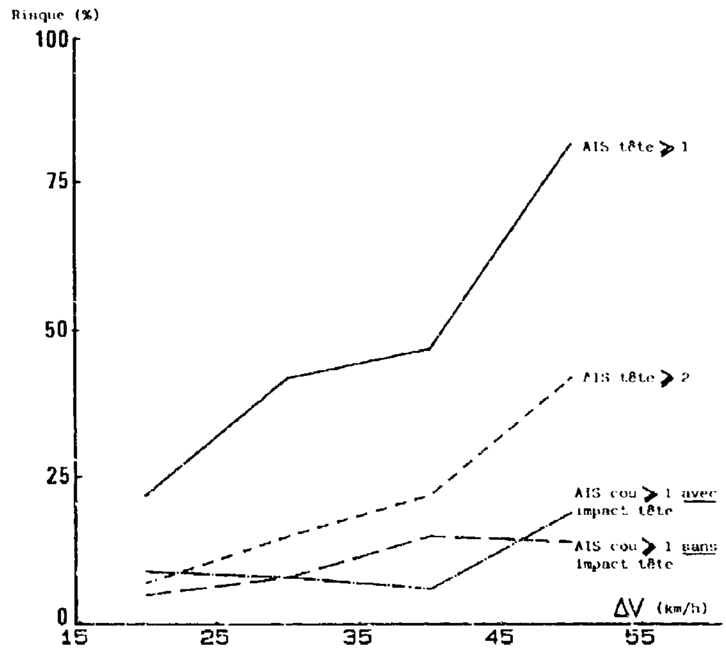


Figure 5. Risque de lésions à la tête et au cou en fonction de la vitesse en choc latéral

	Avec impact de la tête	Sans impact de la tête
Fracture des pédicules	2	2
Fracture du corps	3	2
Fracture épineuse	-	1
Fracture de l'arc postérieur	1	-
Fracture arc postérieur	1	-
Fracture arc postérieur	1	-
Fracture du corps	2	-
Fracture de l'épineuse	-	1
Fracture luxation	1	-
Fracture apophyse transverse	1	-
Fracture épineuse	-	1
Fracture luxation	2	-
TOTAL	14	7

Figure 6. Typologie des lésions cervicales

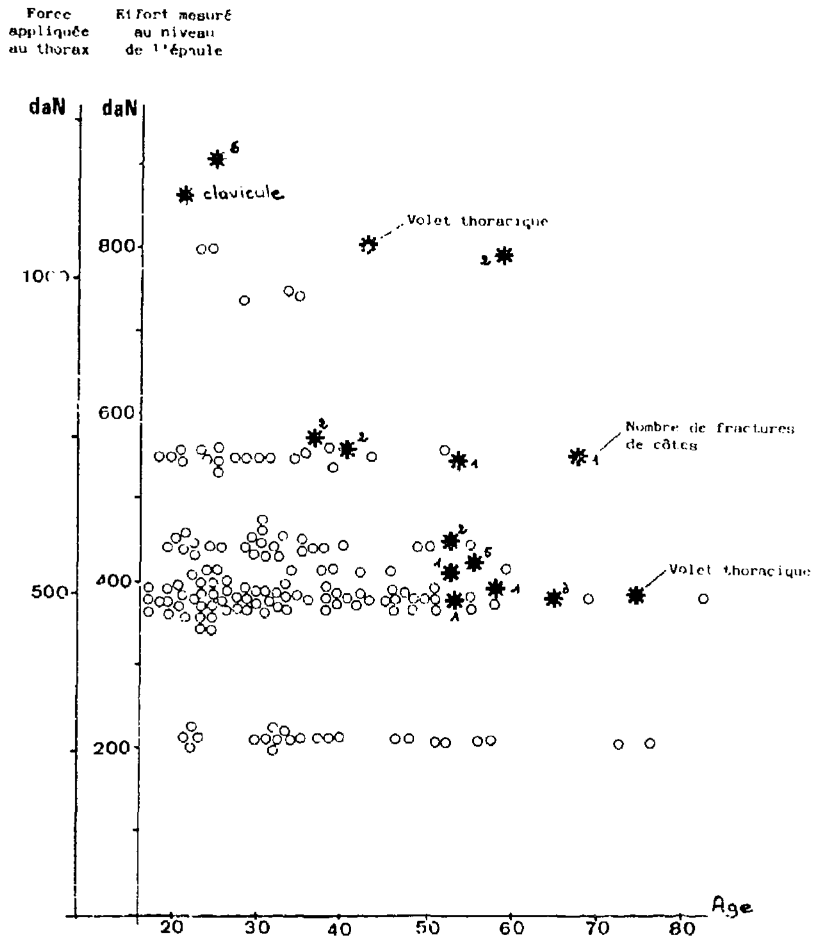


Figure 7. Apparition des fractures de côtes en fonction de l'effort mesuré à l'épaule et de l'âge pour des occupants ceinturés en choc frontal

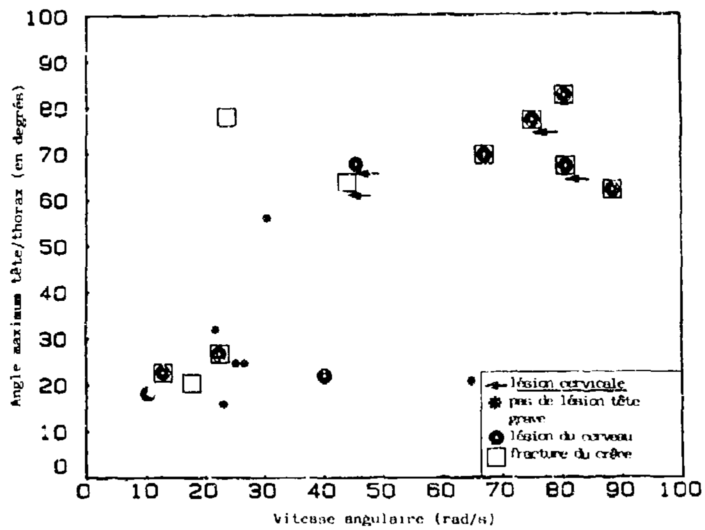


Figure 8. Chutes frontales

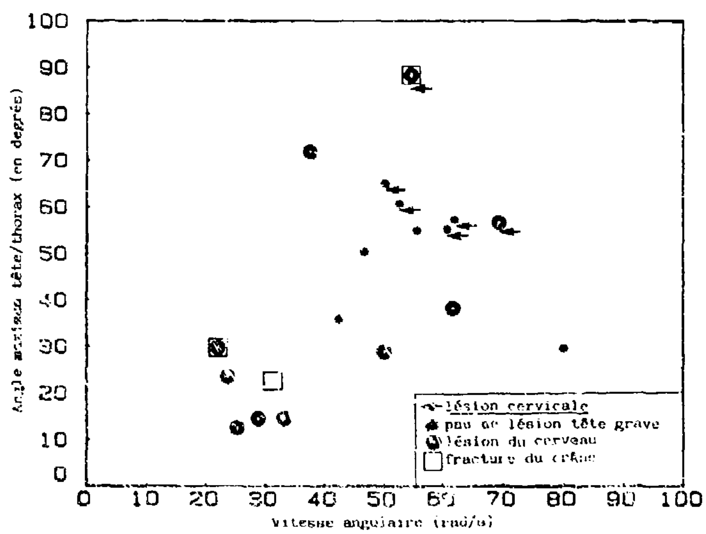
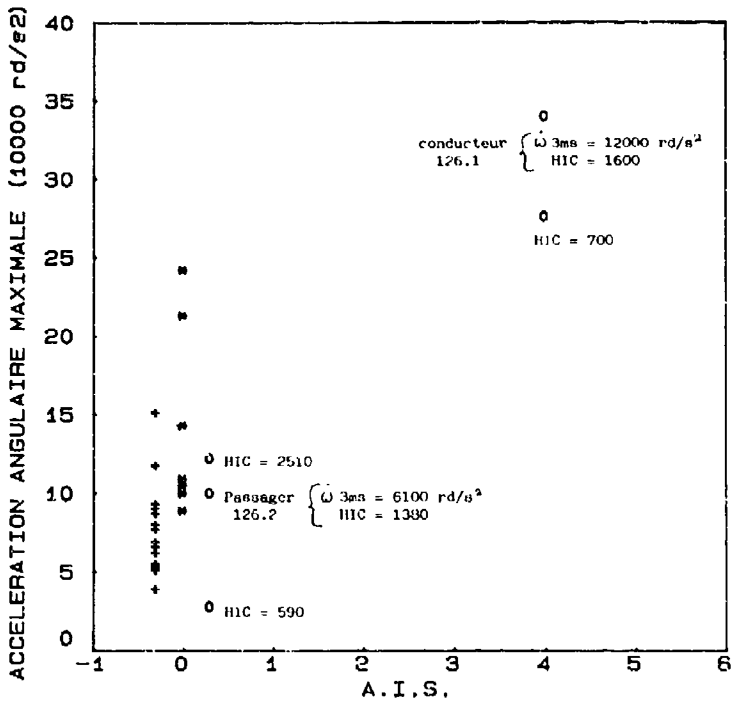


Figure 9. Chutes latérales



O : CADAVRES CEINTURES 3 pts
* : CHUTE FRONTALE (CADAVRE)
+ : IMPACTS BOXEURS

Figure 10. Synthèse de trois sources de données

A Computer Simulation Model For Studying Cervical Spine Injury Prevention

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SUMMARY

Cervical spine fractures, particularly of the burst type due to axial compressive loading of the spine, have been a problem in sport and transportation. Such injuries are usually associated with a head first collision, in which the head strikes a rigid surface (e.g., windscreen, dashboard, etc.), with the neck partially to fully flexed. A computer simulation model has been developed as one means by which protective devices can be evaluated. The model consists of two rigid masses (head and torso), three spring elements (dashpots and non-linear springs) representing the neck and the compliance of the cranium and scalp and three optional spring elements taken to represent the characteristics of protective devices. Simulations using the model, at impact velocities of 1.6 and 3.0 m sec⁻¹, suggest that to maintain cervical spine loads at a non-injurious level (e.g., below 2000N) requires a padding material thickness incompatible with wearing a helmet.

INTRODUCTION

Failure of the cervical spine, with accompanying spinal cord trauma, has been reported to occur in such recreational and sporting activities as diving into shallow water (Tator and Palm, 1981), North American football (Torg, 1985), and ice hockey (Tator and Edmonds, 1984, 1986) and in motor vehicle accidents (Heulke, et.al., 1981). The resulting trauma is often due to axial compressive loading of the cervical spine, with the neck flexed or partially flexed, leading to fracture or fracture dislocation at C_4, C_5, C_6 (McElhaney, et.al., 1979, Mertz, et.al., 1978; Tator and Edmonds, 1984). The neurological deficit imposed by these injuries is significant, with permanent quadriplegia a frequent outcome.

In aviation, the problem of spinal loading in pilot ejections, due to accelerations directed upward through the seat pan or through the sacrum, has been studied intensively. Yoganandan, et al. (1987) have provided a comprehensive review of several spinal models which have been developed to investigate this situation.

Several papers have recently been published (Anderson, 1985, Knudson, et al., 1988; Vanderbeek, 1988; and Anton, 1967) regarding the problems of high-G loading to the necks of aircrew, particularly those flying high performance aircraft. The problems encountered were usually related to soft tissue and included such injuries as muscle and ligamentous strains and their accompanying pain and disability, as well as damage to nerve roots with concomitant paraesthesia and discomfort. Vertebral fracture in the cervical region was not reported. Nevertheless, the potential for vertebral fracture, due to axial compression from crown loading as a result of contact with parts of the aircraft structure continues, and the problems associated with preventing such should be discussed. In this paper a computer simulation model for evaluating the effectiveness of protective padding, placed either on the impact surface or within the crown of a helmet, will be presented.

IMPACT SIMULATIONS

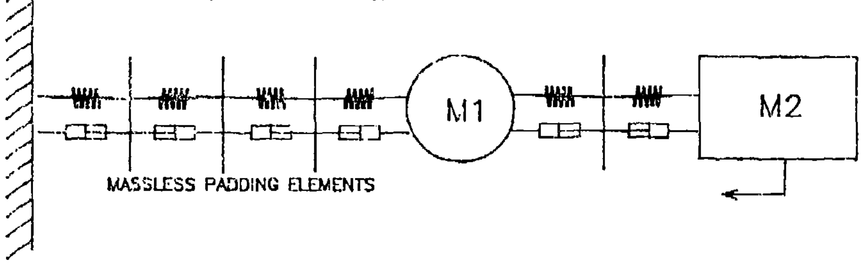
Our concern with axial compressive loading, and the cervical spine fractures that follow, has been directed toward an understanding of the biomechanics involved and with the development and evaluation of strategies for its prevention. Our first efforts involved a number of mechanical simulations using a Hybrid III head and neck, instrumented with a six-axis force transducer (Denton Electronics) and attached to a Hybrid II torso. By mounting the dummy to a pendulum device, it was possible to propel it into a fixed barrier, in free flight, at a known impact velocity. The forces and moments of force, recorded at the transducer, were digitized and stored for analysis using a micro-computer-based analog to digital converter sampling of 2500 Hz. Using this system it was possible to vary the impact environment by fitting the ATD with different helmets and crashing it into different impact surfaces. Results obtained from these simulations (Bishop and Wells, 1986) have been very useful in helping to clarify some of the biomechanical factors associated with this type of injury.

These studies have also been useful in helping to clarify the problems of maintaining low compressive forces on the cervical spine (i.e., prevention) through the use of protective padding, either on the impact surface or within a helmet. Our results have shown that the ATD is decelerated in stages, rather than uniformly. The head, in striking first, comes to rest but the torso continues to move. The neck, trapped between the fixed head and moving torso, must then exert a large reaction force in order to stop the torso. If the energy of the torso is large, the reaction force exerted by the neck will likely exceed that which is tolerable and a vertebral body fracture will result.

COMPUTER SIMULATION MODEL

The methodology used in the mechanical simulations is costly and time intensive. An alternate approach to evaluating surface and/or helmet padding combinations has been developed (Wells, Bishop, and Stephens, 1987). It involves a computer simulation model of the crash situations conducted with the Hybrid III head and neck. Briefly, the simulation model consists of a human representation striking a rigid surface, head first (Figure 1). The body is modelled as two rigid masses, the head (m_1) and the torso (m_2) (i.e., the rest of the body), the neck as two spring elements, and the compliance of the scalp and cranium as a third spring element. Three additional, but optional, spring elements are included to represent the characteristics of protective devices such as an impact surface which deflects, padding applied to the impact surface, or helmet padding, etc. Each spring element includes both a dashpot and a nonlinear spring to better simulate the nonlinear characteristics of biological tissues and polymer-based padding materials.

Fig 1 Computer Simulation Model



The force in any element is determined from:

$$F = k_1 (l_0 - l) + k_2 (l_0 - l)^2 + k_3 \frac{d}{dt} (l_0 - l) \qquad (1)$$

where l_0 is an unloaded element length representing either the padding thickness or the length of the neck. k_1 and k_2 are spring constants, k_3 is a damping constant and l is an instantaneous element length. A variable-step fourth-order Runge-Kutta scheme is used to solve the dynamic equations. The integration routine was driven by an iterative bisection solver which calculated displacements for the massless neck and padding elements. This model is similar to that reported by Sances, et.al. (1984) but includes more neck and padding elements.

The model parameter estimates came from a number of sources. Axial stiffness of the Hybrid III neck was provided from cyclic tests conducted at .13, 1.3 and 13 cm $\cdot$ sec^{-1} in an axial compression mode (McNeice, unpublished data). An estimate of the effective torso mass was made from low speed (.5m.sec^{-1}) ATD impacts and was calculated at 50.0 kg., while a full head mass of 5.0 kg. was used. The compliant properties of the scalp and cranium were lumped and an overall stiffness was found by varying the spring and damping properties until the computer model response matched that of experimental, non-helmeted ATD impacts (Wells, Bishop, and Stephens, 1987).

Simulations are made by inputting the velocity of the collision, the damping and elastic parameters representing the padding material and the material's initial thickness and its thickness at the minimum or "bottomed out" condition. The simulation then returns the forces and displacements of each spring element over the impact.

COMPUTER SIMULATION RUNS

The dynamic response of a polymer-based padding material depends upon two factors, namely the padding's thickness and its stiffness. The computer simulation model can be used to illustrate the relationship between these two factors and the problems of maintaining the compressive forces on the neck to tolerable levels.

A series of simulations was conducted at input velocities of only 1.8 and 3.0 m.sec^{-1} using the properties of a padding material typically found in protective floor matting. The velocity of 1.8 m.sec^{-1} was chosen because it was the velocity used in the mechanical simulations and 3.0 m.sec^{-1} was selected because it represents the threshold of certain injury to the unprotected cervical spine (McElhaney, et.al., 1979). The elastic and damping characteristics of the material were determined by subjecting it to compressive loading up to 5000N, using a Instrom test device, and plotting the corresponding force deflection curve. The parameters to define the curve of best fit were then determined and used in the model. The initial thickness

of the material was 60.0mm, the effective torso mass was 50.0 kg and the head mass was 5.0 kg. As well, the damping constant (k_3) was tailored in order to illustrate its influence on the resulting compressive forces.

RESULTS

An example of the model output is shown in Figure 2. The damping constant (k_3) used was 1.6×10^4 N.sec.m^{-1} and the input velocity was 1.8 m.sec^{-1}. The peak force was 4.8kN and the material deflection output by the model was 8.0mm. At 3.0 m.sec^{-1} the peak compressive force was 8.2kN with a deflection of 13.2mm (Table 1). Thus, in spite of its original thickness (i.e., 60.0 mm) the material responded dynamically as being very stiff, thereby producing very large compressive forces, even at low impact velocities.

Fig 2 Compression Force at 1.8 m/sec for k3=1.6E04 Nsec/m

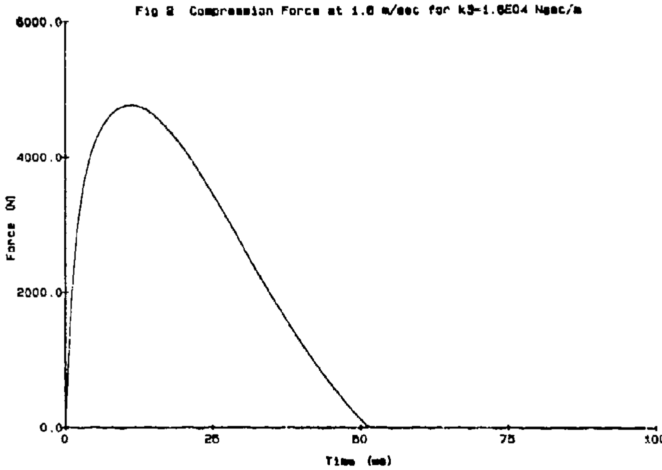


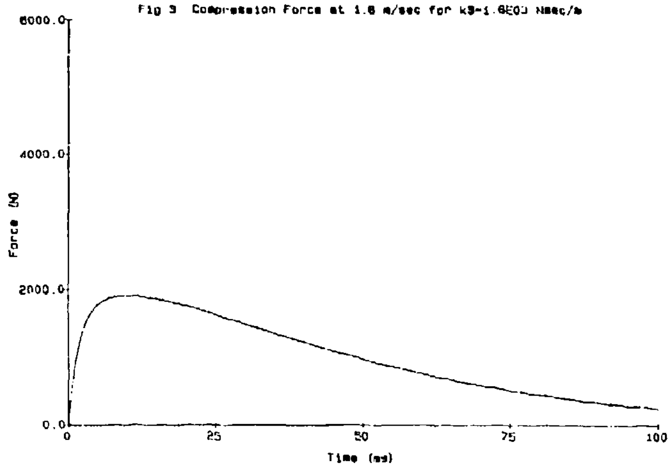
Table 1 illustrates the model response when the damping constant was tailored to reduce the peak compressive force to levels below 2.0kN. The value of 2.0kN was chosen because it represented the lowest load at which whole human spines sustained serious damage during dynamic axial compression (McElhaney, et.al., 1983). With k_3 reduced 10 fold, the result at 1.8m.sec^{-1} was a reduction in the peak compressive force to only 1.9kN (Figure 3), and a material deflection of 59.7mm. To hold the compressive force below 2.0kN, at 3.0m.sec^{-1}, a damping constant of 7.5×10^2 N.sec.m^{-1} was needed and resulted in a material deflection of 174.6mm. Since foam padding materials of this kind bottom out at approximately 70% of their original thickness, this material would have to have an original thickness of more than 250.0mm to be effective.

TABLE 1

Peak Neck Compressive Forces and Material Deflections For Simulations Run At Velocities Of 1.8 and 3.0 m.sec^{-1} With Different Damping Coefficients

	1.8 m.sec^{-1}		3.0 msec^{-1}	
	Peak Force (kN)	Material Deflection (mm)	Peak Force (kN)	Material Deflection (mm)
$k_3 = 1.6 \times 10^4$	4.8	8.0	8.2	13.2
$k_3 = 1.6 \times 10^3$	1.9	59.7	3.2	99.3
$k_3 = 7.5 \times 10^2$	1.2	105.7	2.0	174.7

Fig 3 Compression Force at 1.8 m/sec for k3=1.6E03 Nsec/m



DISCUSSION

The results generated from this simulation model demonstrate some of the difficulty associated with preventing cervical spine failure, due to axial compressive loading, by means of padded helmets or surfaces. To maintain the forces on the neck within tolerable levels, the padding material must be capable of uniformly decelerating the head-neck-torso system. It must be soft enough to deform or deflect over a fairly large distance (6.0 - 17.5 cm) implying an initial thickness of 8.5 - 25.0 cm, and yet be firm enough to dissipate energy without bottoming. Such requirements are incompatible with many sport and recreational activities, and are certainly incompatible with wearing a helmet.

On the other hand the model may be useful in transportation and aviation for the design of vehicular and/or aircraft interiors, where padded surfaces are more readily incorporated. Padding thicknesses of the magnitudes described here, however, are likely to be problematic in high performance aircraft where occupant space is at a premium.

More recently some of the results of our materials testing, and the determination of the parameters to best describe the force-deflection curves so generated, suggest that a fourth-order model may be more appropriate than the second-order model here described. This and other refinements to the model will continue. In the meantime, the most appropriate preventive strategy appears to be the avoidance of situations or behaviours which are likely to induce axial compressive loading.

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BIOFIDELITE DES COUS DE MANNEQUINS AU COURS DES ESSAIS DE CHOCS AUTOMOBILES.

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RESUME

Les mannequins sont largement utilisés dans le cadre des recherches sur la protection du passager d'automobile. Ces mannequins ont été développés sur les bases acquises dans le domaine de la biomécanique du corps humain soumis au choc. Le cou de certains mannequins a été réalisé de façon à présenter un comportement au choc proche de celui de l'être humain.

Dans cette étude nous nous proposons d'analyser, en nous basant sur la bibliographie, la cinématique tridimensionnelle de la tête des mannequins SID, Hybrid III et EUROSID dans le but d'en étudier leur biofidélité.

D'une façon générale, la biofidélité est meilleure pour le choc frontal que pour le choc latéral. Ceci s'explique essentiellement par la non prise en compte de la rotation de la tête autour de son axe vertical dans le développement des mannequins.

La biofidélité et les performances des mannequins reposent sur des bases biomécaniques issues parfois d'essais sur cadavres. Nous tentons ici de nous prononcer sur la validité de tels essais puis analysons les développements futurs possibles.

Nomenclature

- xc, yc, zc : les coordonnées du centre de gravité du chariot dans un repère orthonormé direct avec x dirigé dans le sens postéro-antérieur du sujet.

- xh, yh, zh : les coordonnées du centre de gravité de la tête dans un repère orthonormé direct, avec x dirigé dans le sens postéro-antérieur.

- xT1, yT1, zT1 : les coordonnées de la première vertèbre thoracique dans un repère orthonormé direct, avec x dirigé dans le sens postéro-antérieur.

- Δy = yh - yT1

- θx, θy, θz : les angles de rotation de la tête autour des axes x, y et z passant par T1.

- Fx, Fy, Fz : les résultantes des forces appliquées au niveau de T1, selon x, y, z.

- Mx, My, Mz : Les résultantes des moments appliqués au niveau de T1, selon les axes x, y, z.

- t : La durée du déplacement de la tête.

- . : Symbolise la dérivée par rapport au temps.

1. INTRODUCTION :

L'étude de la biofidélité des mannequins existants, passe par l'analyse détaillée et comparative de la cinématique de la tête dans différentes configurations de chocs. Les mannequins qui font l'objet de cette étude sont ceux qui possèdent un cou développé dans le but de présenter une certaine biofidélité.

Ces mannequins (SID, Hybrid III et EUROSID) sont présentés succinctement dans le premier paragraphe. Dans les deux paragraphes suivants, nous présentons les tests biomécaniques et les essais mannequins de divers auteurs, en distinguant le choc frontal et le choc latéral. Dans le quatrième paragraphe, nous évoquons la question des critères de tolérance et les possibilités de détecter les limites de tolérance au moyen des mannequins. Les deux derniers paragraphes sont consacrés respectivement aux problèmes posés par l'origine et la fiabilité des données biomécaniques, et aux développements futurs des mannequins et des procédures d'essai.

2. PRESENTATION DES MANNEQUINS

HYBRID III

Ce mannequin de choc frontal est le dernier né de plusieurs générations de mannequins développés aux USA dans le cadre des recherches sur la protection des passagers d'automobiles (1,2). Le cou de l'Hybrid III a été conçu dans un souci d'une plus grande biofidélité. Il s'agit d'une structure cylindrique à base d'élastomère et d'anneaux en aluminium. Le comportement dissymétrique en flexion/extension a été obtenu par des entailles aménagées dans la partie antérieure du cou (voir fig. 1). Un câble de précontrainte passe par l'axe du cou et limite ainsi ses déformations en traction tout en facilitant le contrôle de sa loi de comportement. De plus ce mannequin peut être équipé d'un capteur de forces et de moments au niveau de la jonction occipitale.

SID

Le mannequin SID (Side Impact Dummy) destiné à l'étude du corps humain soumis au choc latéral a lui aussi été développé aux USA. Son cou se présente sous la forme d'une simple structure cylindrique dont le comportement en flexion est le même quelle que soit la direction du moment appliqué (voir fig. 2). Ce cou est identique au cou de la Part 572 et n'a pas été réalisé dans le but d'obtenir une bonne biofidélité.

EUROSID

A l'origine de ce mannequin de choc latéral, nous trouvons différents pays de la Communauté Européenne, d'où sa dénomination : European Side Impact Dummy. Son cou a été développé au sein des laboratoires de l'Association Peugeot-Renault (3,4). Il se subdivise en trois parties : la jonction cou/thorax, l'interface tête/cou et la partie centrale réalisée en élastomère (voir fig. 3). Dans cette configuration, le cou permet de moduler la rotation tête/cou et cou/thorax, rend possible une translation pure dans la première partie du mouvement et provoque une torsion du cou combinée avec la flexion latérale.

3. LE CHOC FRONTAL

Cette configuration du choc est la plus fréquente et c'est elle aussi qui a fait l'objet du plus grand nombre de travaux. Les principaux auteurs qui sont à l'origine des données biomécaniques sur le cou en choc frontal sont H.J. MERTZ, R.F. NEATHERY et C.C. CULVER (5, 6). Ces travaux menés sur des volontaires vont être comparés aux tests effectués avec les mannequins HYBRID III (7, 8, 9) et EUROSID (8) par J.K. FOSTER et al, N.M. ALEM et al et F. BENDJELLAL et al.

L'essai de choc frontal consiste à décélérer un chariot sur lequel est assis et ceinturé le sujet d'essai. La description détaillée de la position du sujet est donnée en référence (8). Les valeurs d'entrée qui caractérisent la sévérité du choc subi sont la vitesse et la décélération du chariot ou la décélération de la première vertèbre thoracique. Les paramètres de sortie qui décrivent le comportement du cou sont la position, la vitesse, l'accélération et la position angulaire de la tête, la durée du mouvement de la tête et les éléments de réduction au niveau du cou du torseur des efforts appliqués à la tête (forces et moments transmis par le cou).

Un deuxième type de données, issues d'une autre procédure d'essai, sont les courbes donnant le moment My en fonction de l'angle de flexion ou d'extension $\pm \theta y$. Ces courbes sont obtenues soit par mise en charge de la tête, le sujet restant immobile, soit directement à l'aide de capteurs de moments pour les mannequins.

Outre le capteur "6- axes" donnant les forces et les moments au niveau de la jonction tête/cou les mannequins sont pourvus d'un accéléromètre monoaxial au niveau de la première vertèbre thoracique et d'un triaxial au niveau du centre de gravité de la tête. Enfin l'essai est enregistré par cinématographie rapide ce qui donne accès aux paramètres de positions angulaires.

Les données biomécaniques du tableau 1 sont issues des références (7) (6) et (9). Ces données sont relativement incomplètes dans la mesure où aucun paramètre n'est systématiquement mentionné. De plus, au niveau des efforts appliqués, aucune donnée biomécanique n'est disponible.

Pour les essais menés sur le mannequin HYBRID III, il manque souvent les conditions d'entrée, ce qui empêche toute comparaison entre les différents essais. En référence (8), cependant, nous observons une série d'essais à sévérité croissante avec un enregistrement des divers paramètres. Nous pouvons observer que les angles de flexion maximum enregistrés correspondent aux valeurs données par les volontaires. En référence (9), des tests comparatifs ont montré que l'HYBRID III a un comportement proche de celui du volontaire en ce qui concerne le déplacement postéro-antérieur de la tête et l'accélération de ce mouvement (xh et $\ddot{x}h$).

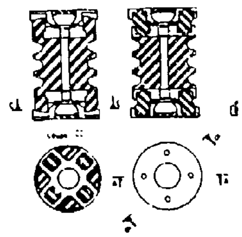
Le mannequin EUROSID a fait l'objet d'un nombre plus limité de tests en configuration de choc frontal. Les observations faites confortent notre attente puisque les valeurs enregistrées s'éloignent notablement de celles émanant des volontaires.

Lorsque la charge est directement appliquée à la tête , la "loi de comportement" du cou peut être approchée par la courbe donnant le moment en fonction de la position angulaire de la tête. Ces courbes $M = f(\theta)$ en flexion et en extension sont données en figure 4 et 5. nous y superposons les résultats relatifs à

Fig. 1 : Cou du mannequin HYBRID III

Fig. 2 : Vue du mannequin SID

Fig. 3a : Vue du mannequin EUROSID



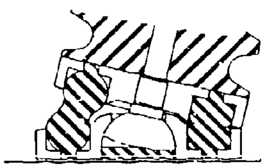


Fig. 3b : Vue en coupe du cou du mannequin EUROSID

l'HYBRID III donnés par FOSTER (7) aux résultats de volontaires rapportés par MERTZ (5) et PATRICK (18) sous la forme de corridors. L'observation que nous pouvons faire est que l'HYBRID II présente une bonne biofédilité quant aux courbes M = f(θ).

Les conclusions que nous pouvons tirer de cette analyse sont qu'il y a d'importantes lacunes au niveau des données biomécaniques et que de toute évidence l'EUROSID est peu biofidèle en configuration de choc frontal.

Pour le mannequin HYBRID III, les performances sont meilleures puisqu'il reproduit convenablement l'angle de flexion θy en test chariot ainsi que les courbes My = f(θy) sans, toutefois, que l'aspect dynamique de ces courbes ne soit spécifié. Relativement aux autres paramètres, tels que les forces et les moments développés au cours d'un essai, il est aujourd'hui difficile d'affirmer que l'HYBRID III présente ou non une bonne biofidélité.

4. LE CHOC LATERAL

C'est à partir des années 1980 que la recherche dans le domaine de la protection des occupants de véhicules a attaché une importance considérable au choc latéral. Les études biomécaniques ont été effectuées d'abord sur des volontaires par EWING et al (10) et par WISMANS et SPENNY (11) puis sur des volontaires et des cadavres par MERTZ (12) et BENDJELLAL et al (13). Ces essais menés parallèlement sur des volontaires et des cadavres ont obligé les chercheurs à distinguer deux niveaux de sévérité des chocs: les chocs modérés (M) et les chocs sévères (S). Les études consacrées au mannequin SID et EUROSID que nous aurons l'occasion d'analyser sont issues des références (6), (13), (14), (15), (16), (17). Au cours d'un test de choc latéral, le sujet est assis et ceinturé sur le chariot qui est alors accéléré lentement puis décéléré selon une loi bien définie. La description détaillée de la procédure d'essai peut être consultée en référence (13). Les paramètres d'entrée et de sortie du système sont les mêmes que pour le choc frontal, à savoir, relatif au chariot, à la première vertèbre thoracique à la tête et aux efforts en présence.

Tout comme pour les sollicitations frontales il existe, pour le choc latéral, des essais basés sur la mise en charge progressive de la tête avec enregistrement simultané du moment et de la rotation de la tête autour de l'axe antéro-postérieur.

L'instrumentation des mannequins de choc latéral consiste simplement en des accéléromètres, l'un monoaxial au niveau de la première vertèbre thoracique, l'autre triaxial au niveau du centre de gravité de la tête. Là encore les angles de rotation de la tête sont donnés par enregistrement cinématographique rapide.

Dans le tableau n°2, nous consignons les résultats qui constituent la base des données biomécaniques et des spécifications ISO issus des références (6), (13), (14) et (18). Nous pouvons noter tout d'abord que les données sont nettement plus complètes que pour le choc frontal. De plus, nous disposons pour les principaux paramètres, de résultats relatifs à deux niveaux de sévérité, M et S, correspondant respectivement aux volontaires et aux cadavres.

Les valeurs relatives aux essais menés sur le mannequin SID nous sont proposées dans les documents référencés (14), (16) et (17). Dans le tableau n°2 nous observons que pour des chocs de sévérité modérée, le mouvement de flexion latérale est convenablement reproduit, mais que l'accélération selon l'axe y et la rotation autour de l'axe z de la tête sont peu ou pas prises en compte. Pour les chocs sévères, des différences avec les valeurs biomécaniques sont plus accentuées encore, puisque l'accélération de la première vertèbre thoracique et dans certains cas l'angle de flexion latérale de la tête ne respectent plus les spécifications ISO. Enfin nous ne disposons d'aucun résultat relatif aux sollicitations transmises par le cou pour la simple raison qu'il n'y a pas de capteurs prévus à cet effet.

Pour l'EUROSID, un grand nombre de résultats sont disponibles et nous présentons au tableau 2, ceux issus des documents (13), (14), (15) et (17). Ce mannequin répond convenablement aux spécifications ISO dans les configurations de chocs modérés, sauf pour ce qui est du déplacement vertical de la tête et de son angle de rotation autour de l'axe z.

Les chocs plus sévères semblent moins bien approcher la "réalité" puisque s'ajoutent aux paramètres défaillants cités ci-dessus, l'accélération selon l'axe y de la tête et de la première vertèbre thoracique.

L'EUROSID n'étant pas pourvu de capteurs d'efforts au niveau du cou, nous ne disposons d'aucun résultat relatif à ces paramètres au cours des tests de choc latéral. C'est certainement ce manque d'instrumentation qui explique également le fait que nous ne disposons d'aucune courbe relative aux mannequins qui puisse être superposée aux courbes moments-angle de flexion latérale rapportée en figure 6.

En conclusion nous pouvons dire que le SID présente une biofidélité assez restreinte et que l'EUROSID a un comportement plus proche des spécifications ISO. Il est à noter cependant que l'angle de rotation autour de l'axe z est mal approché quel que soit le mannequin considéré. Dans le cas de chocs sévères, les données biomécaniques nous semblent peu fiables vu le petit nombre de tests disponibles et les fractures observées sur les cadavres après essai. Ce point sera discuté ultérieurement. Sur le plan dynamique, toute évaluation de la biofidélité des mannequins s'avère impossible puisque contrairement aux cas du choc

Table 1 . Tableau de résultats en choc frontal

Réf 7 : Z.K. FASTER (1977)

Réf 8 : N.M. ALEM (1977)

Réf 6 : Y.W. MELVIN

Réf 9 : F. BENDJELLAL

N° Réf.	7		8			6	9		
Objet	H III	Vol.	H III	H III	H III	Base Bioméca.	Vol.	H III	Euresid
$\dot{X}v$ m/s			[illegible],3	10,5	15,2				
$\ddot{X}v$ G			9,5	10,5	19,4				
$\ddot{X}11$ G							212	170	170
Xh cm			58	58,4	58		20	15	20
Yh cm			47	58	58				
Zh cm			47	44	40		17	7	20
$\dot{X}h$ cm/s			58	58	55				
$\dot{Y}h$ cm/s			58	56	58				
$\dot{Z}h$ cm/s			61	63	56				
$\ddot{X}h$ G			105	102	104		218	218	131
$\ddot{Y}h$ G			133	88	73				
$\ddot{Z}h$ G			112	108	85		+11	-15	-25
$\theta y°$ +	71	64 - 78	74	78	83				
−	82	64 - 85							
T ms			170	170	170		170	170	170
Fx N			107	100	81				
Fz N			83	79	64				
My N.m +	130		102	85	69				
−	37								
M f (θ) +	Oui	Oui				Oui			
−	Oui	Oui				Oui			

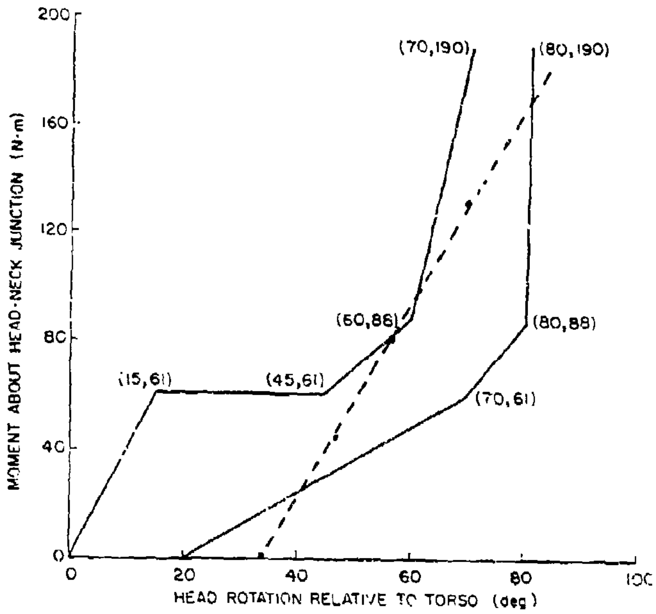


Fig. 4 : Courbes moments de flexion en fonction de la rotation de la tête
———Mertz- - - - - HYBRID III.

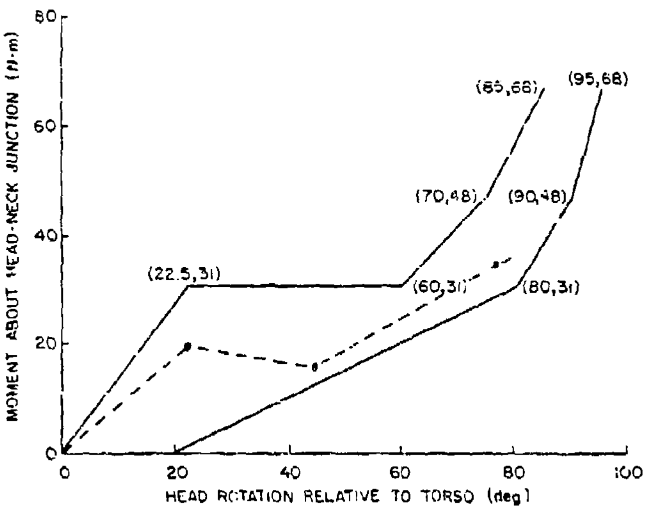


Fig. 5 : Courbes moments d'extension en fonction de la rotation de la tête ———Mertz - - - - -HYBRID III.

N° Réf	6	13				14			15	16	17	
Objet	Vol	Euro	Vol	Euro	Cad	ISO	Euro	Sid	Euro	Sid	Euro	Sid
$\dot{y}c$ m/s		6,2	6,4	6	5,1	5,9-6,2			9,3	9,1		
$\ddot{y}c$ G		7,2	7	13,8	12,2	S10-14						
						M 6- 9	7	7	15	15	12	12
yT1 mm		45	69	52,5	67	40-63	40	75				
$\ddot{y}$T1 G		13,1	17,3	19,8	20	S17-23						
						M12-18	12	18	43	55	30	30
Δy mm						130-162	110	170				
yh mm		164	153	191	294	185-226			252	305		
zh mm		29	89	60	79	61-96						
$\ddot{y}h$ G		9	12,5	9,6	36	S25-47						
						M 8-11	9	13	89	32	X	X
$\ddot{z}h$ G						8-10	10	7				
θx °		54	50	72	78	S62-75						
						M44-59	53	50	122	90	75	72
θz °		22	50	48	42	S62-75						
						M32-45			2	0		
t ms						159-175	110	110				
Fx N						325-375						
Fy N						750-850						
Fz N						350-400						
Mx Nm						40-50						
My Nm						20-30						
Mz Nm						15-20						
M = f(σ)	oui	non				non						

Table 2 : Tableau de résultats en choc latéral

Réf 6 : Y.W. MELVIN (1985)
Réf 13 : F.B. BENDJELLAL (1986)
Réf 14 : ISO Doc 213 (1988)
Réf 15 : ISO Doc 199 (1988)
Réf 16 : ISO Doc 211 (1988)
Réf 17 : ISO Doc 214 (1988)

frontal où nous disposions des efforts dans le cou des mannequins mais pas dans les sujets anatomiques, ici nous avons les efforts transmis par le cou des sujets anatomiques mais pas des mannequins.

5. CRITERES DE TOLERANCES

Les recherches sur les mécanismes de lésions cervicales ont abouti à des critères de tolérances au niveau du cou, qui sont exprimés en termes dynamiques et cinématiques (6,1920). Au niveau des mannequins ces critères deviennent des limites ultimes à ne pas dépasser et les grandeurs physiques qui définissent ces critères doivent constituer les sorties ou les résultats d'un essai de choc. Les critères actuellement retenus pour les lésions cervicales sont les moments maximum ($Mx = \pm 120$ Nm, $My = + 190$ Nm, $My = - 57$ N.m) et les forces maximales transmises par le cou ($Fx = Fy = 1\ 000$ à $3\ 000$ N, $Fz = + 3500$N). Au niveau cinématique, les limites suivantes ont été fixées : $\theta x = \pm 50°$ $\theta y = 70°$, $\theta y = - 90°$, $\theta z = \pm 70°$.

Le premier point que nous pouvons regretter est que l'instrumentation actuelle des mannequins ne permet pas de mesurer ces paramètres et que, sur les sujets anatomiques, ces valeurs ne sont pas systématiquement calculées à partir des données cinématographiques.

Du point de vue des données biomécaniques il faut signaler que les paramètres fixant les critères ne sont pas indépendants, puisque le dépassement des limites cinématiques entraine le dépassement des limites dynamiques et vice versa.

Il faut remarquer ensuite qu'aucune limite de tolérance n'est donnée relativement à Mz et que, d'une façon générale toutes les limites sont données sans que soient précisées les conditions de vitesses et d'accélérations des mouvements étudiés.

Enfin, les critères proposés ne prennent pas en compte le risque de lésions provoqués dans des configurations de flexion rotation ($\theta x + \theta z$) qui sont pourtant effectives en cas de choc latéral.

6. LIMITES DES DONNEES BIOMECANIQUES

Les essais de chocs utilisant des volontaires pour l'étude du comportement dynamique du corps humain resteront, par nature même, des éléments de base fondamentaux dans les données biomécaniques servant de référence à l'élaboration des mannequins. Il ne faut pas oublier cependant que ces essais ne nous donneront pas de résultats pour les chocs sévères et ne nous permettront pas d'accéder aux critères de tolérances. De plus nous pouvons reprocher à ces tests que le sujet soumis au test n'est pas représentatif de la population soumise au risque et qu'il s'attend à subir un choc, ce qui n'est pas toujours le cas dans la réalité.

L'utilisation de cadavres pour la détermination des réponses au choc du corps humain repose sur des hypothèses qui ne sont pas toujours satisfaites.

Il est supposé que la structure squelettique du cadavre ne se trouve pas changée après la mort. Vu la nature relativement inerte du matériau os, cette hypothèse semble justifiée.

Il n'en va pas de même, par contre pour la tonicité musculaire et pour les actions ligamentaires qui peuvent modifier notablement les conditions initiales de l'essai. Pour des impulsions brèves, le système neuromusculaire n'est certes pas capable de recruter les composants actifs des muscles. Ce sont alors les composants passifs et les ligaments qui interviennent et cette action est certainement différente sur le vivant que sur le mort. Cette différence dans les lois de comportement des articulations a une influence significative sur la réponse des sujets, notamment en termes de valeurs extrêmes des paramètres dynamiques.

De plus, les cadavres testés en choc latéral présentent souvent des lésions graves soit au niveau du cou soit au niveau de la zône d'impact, ce qui risque également d'altérer les résultats.

Enfin et pour terminer, le cadavre ne pourra jamais être testé qu'une seule fois et la dispersion des propriétés mécaniques des squelettes des sujets restera toujours importante et difficile à intégrer dans le traitement des résultats.

Cet ensemble d'inconvénients que présentent les tests effectués sur des cadavres doit nous amener à nous interroger sur la validité des données biomécaniques basées sur ce type d'essais.

7. ORIENTATIONS FUTURES

Ce paragraphe, consacré aux orientations envisageables pour l'amélioration de la biofidélité des mannequins, est subdivisé en trois parties consacrées respectivement à l'aspect biomécanique, au problème des mannequins et aux procédures d'essais. Avant toute réflexion il faut également avoir à l'esprit le but à atteindre, c'est à dire l'utilisation du mannequin.

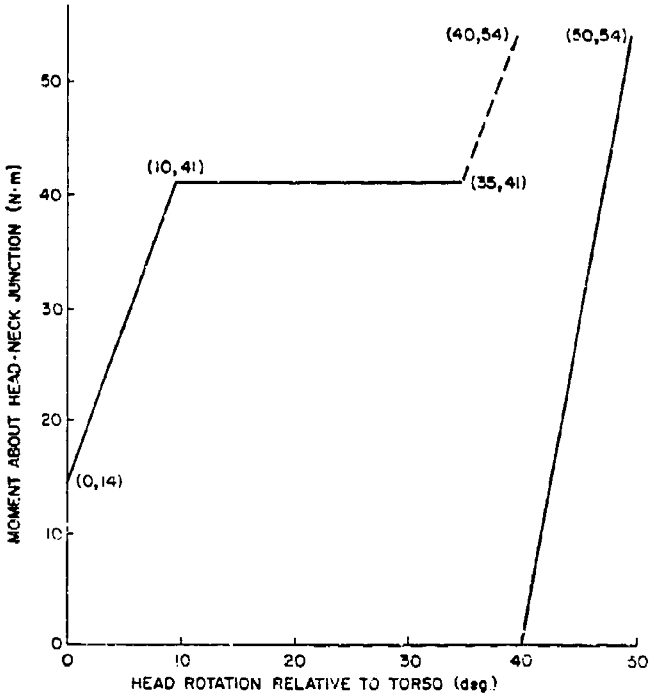


Fig. 6 : Courbes moments en fonction de la rotation latérale de la tête (Patrick et al).

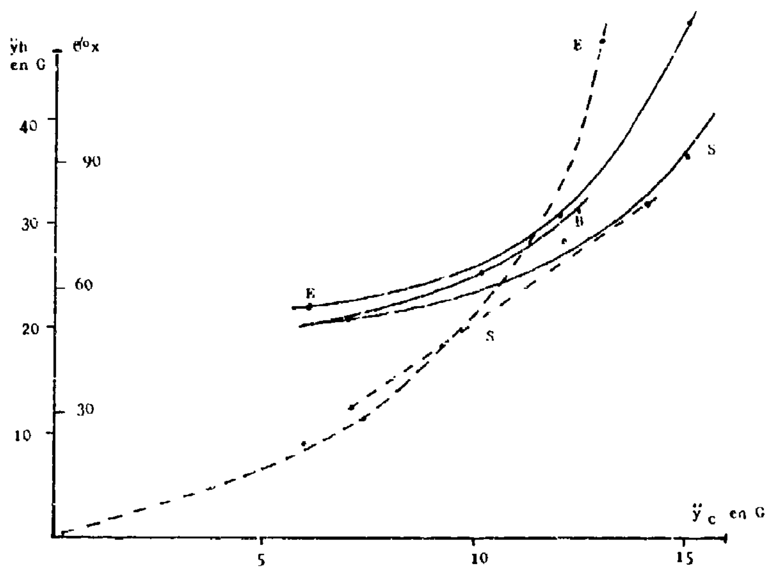


Fig. 7 : Evolution de la rotation latérale θx et de l'accélération de la tête ÿh en fonction de la sévérité du choc ——— θx - - - - - - ÿh.

La première lacune constatée dans les données biomécaniques se situe au niveau de la détermination des éléments de réduction du torseur des efforts appliqués à la tête au cours d'un essai de choc. Ces données, en termes de forces et de moments, pourraient être calculées aisément à partir de l'enregistrement cinématographique puis complétées par les courbes moments en fonction des angles de rotation de la tête qui constituent certainement l'une des meilleures caractéristiques du cou.

Dans un deuxième temps le problème de la validité des tests effectués sur cadavres devra être abordé à travers des chocs à sévérité modérée conduit parallèlement sur des cadavres et des volontaires. Le comportement du cadavre pourra également être pondéré en fonction d'un paramètre relatif à l'action passive des muscles et des ligaments issus de modèles mathématiques des lois de comportement de ces tissus.

Si le mannequin HYBRID III satisfait relativement bien aux conditions à remplir par un mannequin de choc frontal, les défaillances en configuration de choc latéral restent importantes pour les mannequins développés à cet effet.

Le plus biofidèle d'entre eux, l'EUROSID, reproduit effectivement très mal la rotation de la tête autour de son axe vertical que l'on peut observer sur les sujets anatomiques. Par ailleurs, une évolution de ce mannequin est en cours et prévoira des capteurs d'efforts au niveau de son cou, ce qui donnera accès aux forces et aux moments transmis par cette articulation.

Les recherches que nous pourrions envisager quant aux procédures de test de biofidélité des cous s'orientent vers l'étude de la loi de comportement de l'articulation du cou avec la définition des paramètres caractéristiques du système.

Une telle étude devra reposée sur une procédure d'essai simple, précise, répétable, facile à mettre en oeuvre et représentative de la configuration de choc étudiée. Pour le cou, une telle approche consisterait à mettre en charge la tête de façon totalement contrôlée et d'établir les relations qui lient la cinématique aux paramètres dynamiques du système. Ces fonctions, telle par exemple l'évolution du moment en fonction de l'angle à accélération angulaire variable, constitueraient les critères de quantification de la biofidélité d'un substitut du corps humain.

Un exemple peu précis, basé sur les résultats du tableau n°2, est donné en figure 7. Nous pouvons en conclure que le SID et l'EUROSID présentent une bonne biofidélité pour le paramètre θx au cours des tests chariots, si l'on suppose que les masses en mouvement sont proches.

8. CONCLUSION

Dans une certaine mesure, la biofidélité des mannequins disponibles pour les essais de chocs automobile est acceptable et ces substituts du corps humain sont largement utilisés pour l'évaluation de la protection des passagers.

Les points sombres de leur biofidélité se situent bien sûr au niveau de la rotation de la tête autour de son axe vertical en configuration de choc latéral mais d'importantes lacunes subsistent également dans les données biomécaniques relatives aux sujets anatomiques.

L'appréciation de la biofidélité des mannequins basée sur des essais lourds, complexes et avec un grand nombre de paramètres difficiles à contrôler et pas toujours rapportés devra dans certains cas s'orienter vers des procédures d'essais plus légères, plus systématiques et tout aussi proches des conditions réelles de chocs.

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OMNI-DIRECTIONAL HUMAN HEAD-NECK RESPONSE

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ABSTRACT

The Naval Biodynamics Laboratory (NBDL) in New Orleans has conducted an extensive research program over the past years to determine the head-neck response of volunteer subjects to impact acceleration. These subjects were exposed to impacts in frontal, lateral and oblique directions. An analysis of a limited number of frontal and lateral tests from a test series conducted in the late seventies with two subjects showed that the observed head-neck dynamics can be described by means of a relatively simple 2-pivot analog system.

The present study extends this analysis to a more recent NBDL test program with 16 human subjects. The database consists of 119 frontal, 72 lateral and 62 oblique tests. The research methodology used for this analysis includes a detailed description of three-dimentional kinematics as well as load calculations near T1 and the occipital condyles. A description of this research methodology and a summary of the major test results is presented. Special attention is given to the influence of impact severity and impact direction on the head-neck dynamics. It is shown that a similar analog system as proposed earlier for frontal and lateral impacts, is suitable for all impact directions. Geometrical properties of this analog have been determined by means of newly developed numerical techniques rather than through the graphical techniques that were used earlier. Findings of this analysis are discussed in view of future omni-directional mechanical neck developments.

This study has been presented earlier at the 1986 STAPP Car Crash Conference
(SAE paper 861893)

MEASUREMENT TECHNIQUES, EVALUATION CRITERIA AND INJURY PROBABILITY ASSESSMENT METHODOLOGIES DEVELOPED FOR NAVY EJECTION AND CRASHWORTHY SEAT EVALUATIONS

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ABSTRACT

Head and neck injuries have been of particular concern to Navy researchers and have initiated extensive programs to address head and neck response of both live human subjects and human analogues to crash impact forces. This concern has been somewhat heightened, as of late, by the apparently conflicting operational requirements of having canopy penetration as the principal method of ejection in several aircraft prototypes, coupled to the requirement of introducing night vision capability in attack aircraft. The latter will most probably lead to increased helmet volume, and possibly weight, which increases the probability of helmet canopy acrylic interaction during canopy penetration. Increased helmet weight and center of gravity shifts, together with altered helmet to head coupling, will certainly change head and neck response to even presumably "safe" exposure levels.

In order to adequately parameterize head and neck response and relate the gathered data to known living human subject and cadaver data, both inertial response (linear and angular accelerations) and load data (extension, compression, shear and torques) must be obtained at well defined, anatomically correlatable points. A modified Hybrid III type head and neck complex was developed, ballasted to be in compliance with Navy generated head and neck mass distribution parameters, and fully instrumented (inertial and load transducers) at the head center of gravity (C.G.), occipital condyles, and base of neck (approximately T1 level). The fully instrumented head and neck system has been utilized to evaluate various helmet configurations and the effect on head and neck response with changes in helmet weight and geometry. Additionally, neck extension, compression, shear forces and torques were obtained during dynamic ejection tests ranging from 0/0 to 720 KEAS. At the higher airspeeds, the effects of aerodynamic lift can be clearly identified on the monitored neck compression - tension values. With such data, injury modalities and probabilities can be addressed in considerably greater detail than the present norm and the effectiveness of protective equipment established.

INTRODUCTION

With the advent of minaturized, sturdy, and reliable six axis load cells, test manikin segment responses can be parameterized beyond the traditional acceleration measurements obtained. Additionally, load measurements, although directly correlatable to acceleration, give greater insight into the severity of response, since compression and shear forces, together with the generated moments about selected anatomical points, can be easily visualized in terms of biodynamic injury.

A series of horizontal accelerator and ejection tower tests have been completed to establish baseline values for these measures under a variety of initial position and restraint configurations. Additionally, for the head and neck system, the sensitivity of the resulting measured values to changes in head weight and center of gravity was also established. (1) This data can be interpreted as the baseline values against which new helmet configurations (such as night vision) will be compared and from which relative safety assessments can be made. This baseline data is the first of its kind and demonstrates a significantly improved capability to analyze and quantify canopy penetration severity and helmet lift forces during high "Q" escape.

In previous programs conducted at NAVAIRDEVCEN, state-of-the-art manikins (HYBRID II, HYBRID III, VIP) were comparatively tested under identical experimental protocols. From these tests, recommended torso instrumentation was identified (6 linear and angular acceleration measures at the head, upper thorax and pelvis; 6 axis load cells at the head/neck, neck/thorax and lumbar spine/pelvis junction for a total of 36 torso channels) and the HYBRID III type manikins, together with the associated sensor instrumentation system and solid state data acquisition system (96 channels; 4MBytes of memory) were extensively tested both at NAVAIRDEVCEN facilities (ejection tower, horizontal accelerator), up to 48 G, as well as flight tested at the Supersonic Naval Ordnance Research Track (China Lake), up to 620 KEAS (2, 3, 4, 5). This first generation of "biofidelic" manikins designated as BFM1, are extensively employed at NAVAIRDEVCEN in the evaluation of both ejection and crash worthy seating systems and improvements in both the head and neck systems, as well as the pelvic area are well underway. BFM1 is presently the standard test article for both the ejection tower and horizontal accelerator and is instrumented as per Figure 1.

With this sensor arrangement, the dynamic response of the respective segments of interest is completely defined at locations corresponding to high incidence of injury. Additionally, transmission of forces up the spine, eminating from the pelvis and

terminating at the head, can also be used to evaluate seat-man interaction and restraint efficacy, since the three - dimensional acceleration time history and relative displacement of the individual segments is known or can be calculated. In most applications, data has to be translated to other points than those directly instrumented, either to obtain estimates of responses that cannot be directly monitored or as an independent measure used as a cross check to establish accuracy of monitored values. In manikin tests, as an example, one might want to calculate accelerations or torques a the head pivot and only the head center of gravity (CG) acceleration measurements are available. Similarly, in the case of human runs, monitored "head" accelerations may have to be translated from the instrumentation mounting platforms to the head CG or the occipital condyles, since one cannot directly monitor responses at those locations. Body referenced inertial acceleration of a point fixed in a moving rigid body is expressed as a function of tri-axial accelerations, angular rates and angular accelerations (yaw, pitch, roll) about the respective orthogonal axes. The sensors selected to measure these responses have been described by Frisch (1983) and consist of subminiature linear accelerometers, angular rate sensors and angular accelerometers. They are state-of-the-art and have a successful history of utilization in both human biodynamic research and escape system and crashworthy seat testing. Their miniaturized designs are ideal for manikin applications since they are light weight and easily incorporated into the manikin segments for response analysis. All are commercially available, off the shelf items, which have been proven to be dependable, require little or no maintenance and are in keeping with performance capabilities of the data acquisition, storage, and telemetry systems. In addition to the inertial instrumentation, six axis load cells are also incorporated at the head/neck, neck/thorax and lumbar spine/pelvis junctions. Figure 2 demonstrates the monitored data in the head/neck area. The inertial instrumentation at the manikin head CG (shown on the left) provides the angular and linear acceleration components about the head coordinate system axes. With this information, and knowing the relative location of the head pivot, torques about the atlanto-occipital (primarily a hinge joint) and atlanto-axial junction (primarily a rotating joint about the odontoid process) can be estimated. The six axis load cells, measuring flexion, extension, lateral bending moments, shear forces, as well as axial and lateral compression and extension loads, provide independent measures of the calculated values. Similar arguments can be made for the neck/thorax (N) and lumbar pelvis junctions.

EJECTION TESTS

It is unusual to instrument the head of the standard ejection test dummies (GARD-CG), partially due to the hinged nature of the neck pivot which restricts head and neck motion to the mid-saggital plane. Traditional dummy thoracic instrumentation usually consists of monitoring the orthogonal linear acceleration components, together with the respective angular rates about these axes (yaw, pitch, roll). The new proposed standard test manikin (as per Figure 3) significantly increases the instrumentation requirements but also eases data analysis and renders results correlatable to known injury mechanisms.

Head and neck data from a dual zero-zero (zero airspeed-zero altitude) test is shown in figure 4. If one had to rely solely on the traditionally monitored inertial data (Figure 4 top) only a somewhat murky analysis would be possible. One notes that the pitch rate for the forward dummy is considerably different from that of the aft position (see event 1 at approximately 175ms). Similar fluctuations are evident in the Gx and Gz accelerations. From this data, by itself, it would be difficult to ascertain whether this phenomena is due to loose restraint or dummy-crewstation impact. Film analysis of this dual ejection, employing canopy fragilization, indicated that in the forward crewstation location, the helmet did not clear the opening created and this contact imparted significant pitch to the head and neck system. When the neck joint stop was reached, the entire dummy torso pitched forward and the head broke out additional canopy acrylic, enlarging the opening originally created. The opening for the aft location was large enough for the dummy-seat combination to pass through. From the acceleration data, the severity of the head-canopy acrylic impact is difficult to parameterize, as is the assessment of injury probability. The neck load analysis, based on load data monitored at the base of the neck (T1), clarifies the situation and eases interpretation (Figure 4 bottom). At time of head-canopy impact, axial compressive forces reach 900 lbs.; three times the magnitude one would expect solely from the ejection forces (aft dummy). The moment about the pitch axis demonstrates the same order of magnitude difference between the two ejection locations. Form the load data, severe neck injury is highly probable.

The same insight gained is also evident in a dynamic ejection (499 KEAS) conducted with the instrumented manikin (Figure 5). We note that both manikins in this dual ejection entered the windstream at time period 1 and considerable perturbation is evident in the subsequent manikin -Gx acceleration profile. Peak manikin deceleration occurs at about 500ms and is extremely reprod ible in both manikins (25-27 G's). Monitored dummy neck loads mirror this scen . The polarity of the neck loads are reversed (plus should be minus and minus sl l be plus) but as can be seen (Figure 5 - middle) the compressive forces during the catapult phase reach about 300 lbs at approximately 125ms. Subsequently, both manikins enter the windstream; the heads are pushed back against the headbox and lift forces on the helmeted head put the neck into extension (axial) reaching approximately 400 lbs. During time period 2, permanent off-set in dummy neck loads for the forward location is evident. This shift of approximately 600 lbs is an anomaly of the data and does not reflect an actual change in

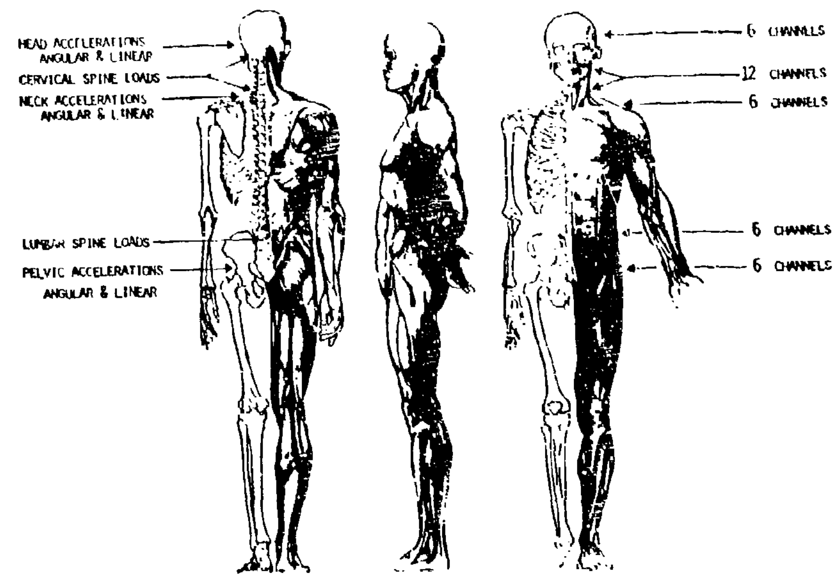

Figure 1. Standardized Biofidelity Manikin Instrumentation Consisting of Inertial and Load Measurements.

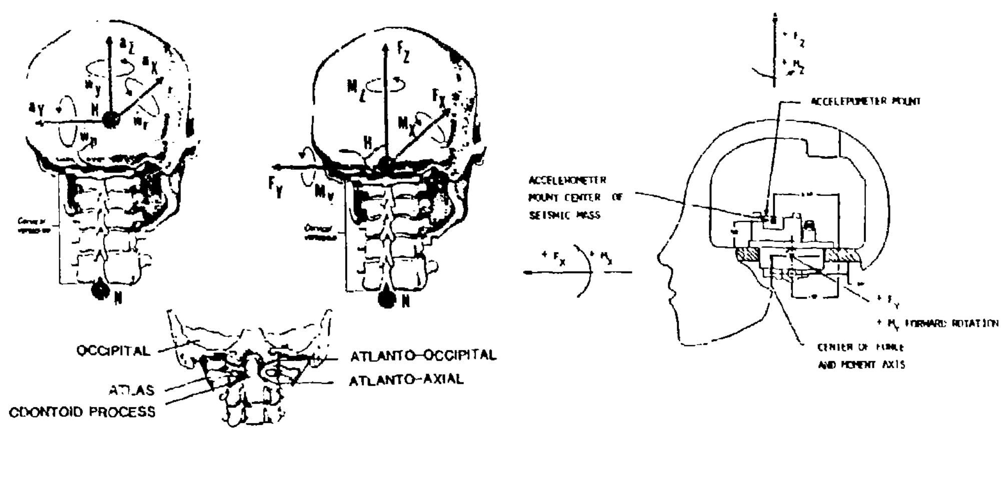

Figure 2. Monitored Head Response Parameters Including Linear and Angular Accelerations About Head Anatomical Coordinate System and Forces and Moments About Occipital Condyles.

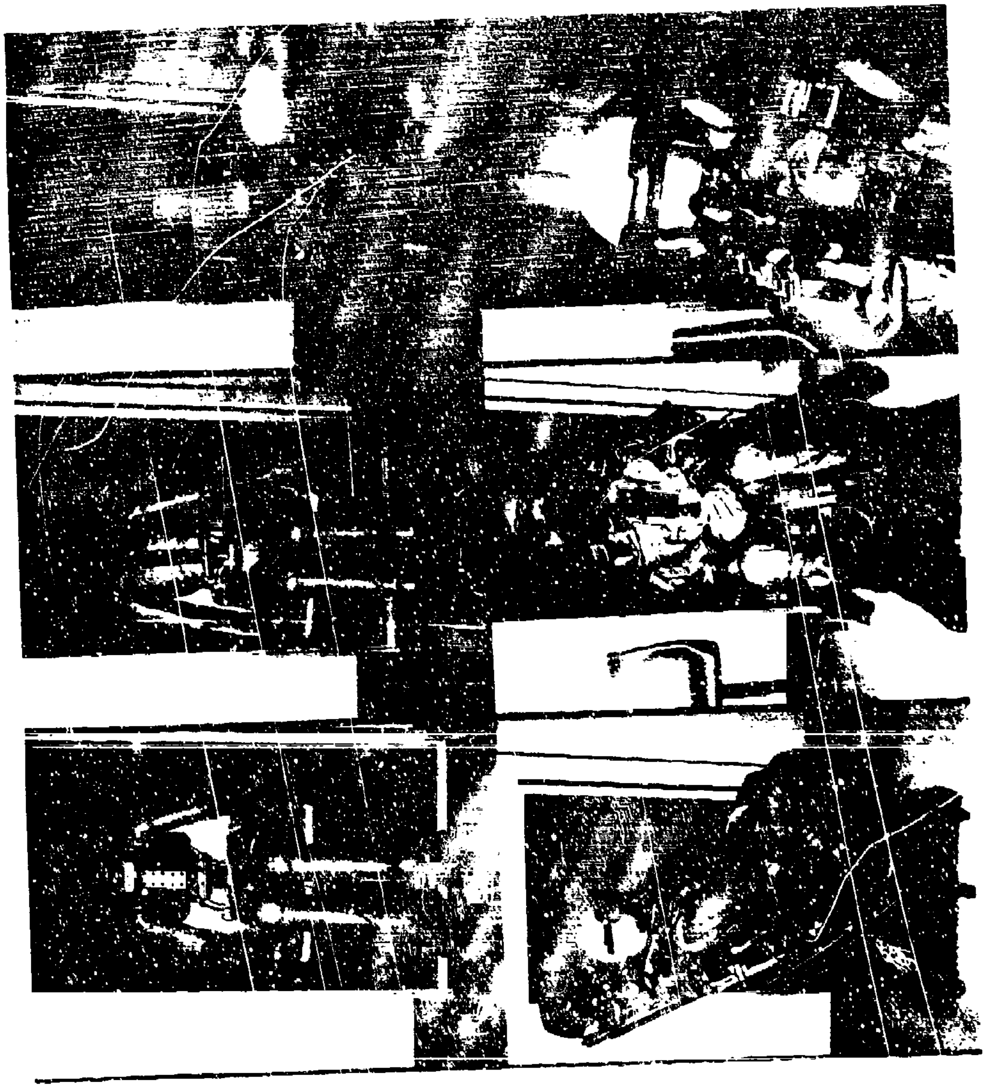

Figure 3. Fully Instrumented Manikin Preparation For Ejection Seat Testing. Note Deformable Cervical and Lumbar Spine Systems.

magnitude. The indicated neck load values never do return to zero but remain off-set at 600 lbs. Assuming this permanent shift in baseline, the relative change in neck extension loads subsequent to time period 2 (400ms) agrees well with that monitored for the aft manikin (400 lbs.). It would appear that this presumed lift force (400 lbs.) accounts for the persistent helmet loss at high airspeeds and confirms a long suspected injury modality of the cervical spine due to neck tension.

The most convincing example of the load cell-linear acceleration instrumentation combination and its utility in analyzing occupant response during ejection is in the data obtained from the transonic (720 KEAS) test conducted with a fully instrumented manikin. As will be noted from Figures 6 and 7, the monitored seat accelerations (Z and X) are consistent with expectations and somewhat higher than normally monitored during seat qualification tests. During the catapult phase of ejection (from approx. 150 ms to 275 ms) the seat attains approximately 15G_z (Figure 6A) resulting in a lumbar load of 725 lbs (Figure 6B) as measured at the pelvic-flexible lumbar spine interface (as per Figure 1). Neck loads (monitored at the head-neck junction representing the occipital condyles) also demonstrate a compressive force of approximately 85 lbs during this time interval. At approximately 300 ms, neck forces are reversed (from compression to tension) achieving neck tension loads in excess of -500 lbs (Figure 6C). The initial interpretation of this data could easily be that the exhibited neck tension is due to aerodynamic lift created by the airflow over the helmeted head. The 500 lbs tension monitored is in good agreement with the 400 lbs obtained in the 499 KEAS test in Figure 5. Both high speed ejection tests indicate that the neck tensions monitored are approaching assumed human tolerance levels (-550 lbs, ref. 6) and these results appear to verify the postulated injury mechanisms associated with the aerodynamic lift characteristics of flight helmets. Unfortunately, the rest of the gathered data does not support this hypothesis. Seat-catapult separation occurs at approximately 300 ms (Figure 7A), subsequent to which the seat undergoes considerable anterior-posterior deceleration (in excess of -20 G_x) and demonstrates a considerable increase in monitored inferior-superior acceleration (approximately 35 G_z - Figure 6A). This latter value is in excess of that anticipated due to seat rocket ignition and indicates, together with the lower then expected monitored G_x values that the seat pitched backward; a fact confirmed by the film coverage. The results of this seat G_z acceleration is clearly seen in the monitored manikin pelvic loads (Figure 6B). Consequently, after catapult separation, one has the situation where the lumbar spine goes into compression (approximately 1800 lbs), whereas the cervical spine is forced into tension (approximately - 500 lbs).

Further film analysis indicated that the helmet came off the head while the seat was still on the rails, and consequently the maximum monitored neck tension forces could not be solely due to aerodynamic lift but were in fact the result of the airstream coming up the manikin chest cavity (since the seat had reclined backwards) and interacting with the chin. Since the upper torso of the manikin is well restrained, this windstream-chin interaction violently rotates the head backward, forcing it against the headbox, which precludes any further head rotation (see Figure 7C for head-headbox impact). Further exertion of force on the underside of the chin puts the neck into tension as seen in Figure 7B. The important thing to note is that in such situations (seat pitching aft), changing the aerodynamic lift properties of the head and neck system may not significantly affect neck tension experienced.

CONCLUSION AND RECOMMENDATIONS

From the data presented, the overwhelming advantages of the instrumentation scheme of Figure 2 are quite clear. This instrumentation scheme will be the new standard for ejection tower tests conducted at NAVAIRDEVCEN. At present, hardware to support this instrumentation configuration is being purchased and data acquisition and storage capabilities are being expanded. This is anticipated to be completed shortly and all future testing will utilize the BFM1 manikin. BFM1 is also being proposed as the standard dynamic ejection test manikin and has been subjected to ejection tests ranging from zero-zero to 720 KEAS. Programs such as Night Vision Goggles (INVS) and 21st Century Helmet have committed both resources and indicated intent to utilize this manikin in their qualification programs. It is anticipated that other programs will follow.

The utilization of both inertial and load sensors greatly enhances the parameterization of the manikin response to various acceleration scenarios and enables robust analyses to be conducted and injury mechanisms and probabilities to be identified. The load cells themselves, being integral structural members of the manikin, have been shown to be reliable and able to withstand the most severe escape conditions and crashworthy seat environments without damage. Their full integration into the manikin anatomical segments has been accomplished without compromising manikin dimensions or performance characteristics.

Work is presently underway within the Navy on BFM2. This effort maintains both the sensor configuration as well as the operating characteristics of the developed data acquisition and storage system (DASS), although some repackaging is being undertaken to fully integrate portions of the DASS into a redesigned, anatomically representative, pelvis. Additionally, the head and neck system has also been modified to improve biofidelity. Softening of the neck column was indicated when manikin response was

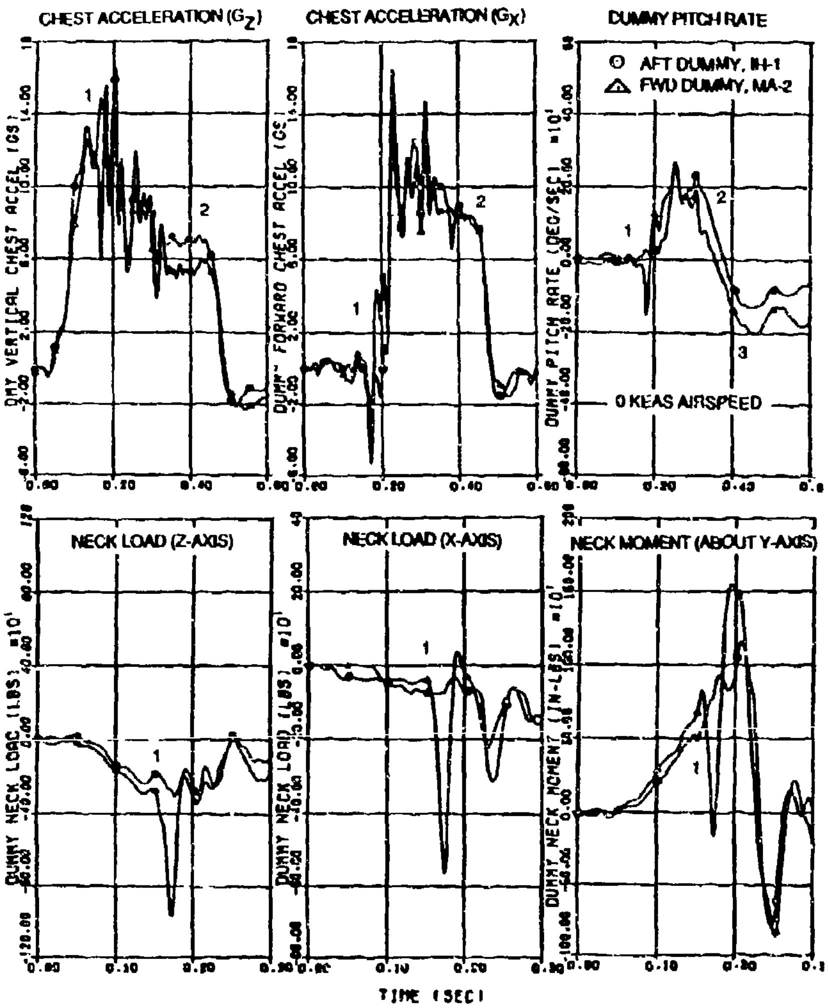

Figure 4. Monitored Manikin Response Parameters From Zero-Zero Test. Note Clear Evidence of Head - Canopy Acrylic Impact in Neck Load Channels.

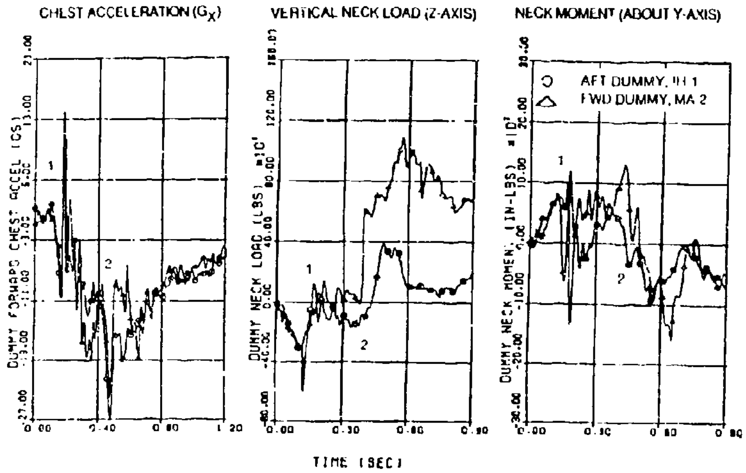

Figure 5. Manikin Head and Neck Response to Windblast Conditions (500 KEAS).

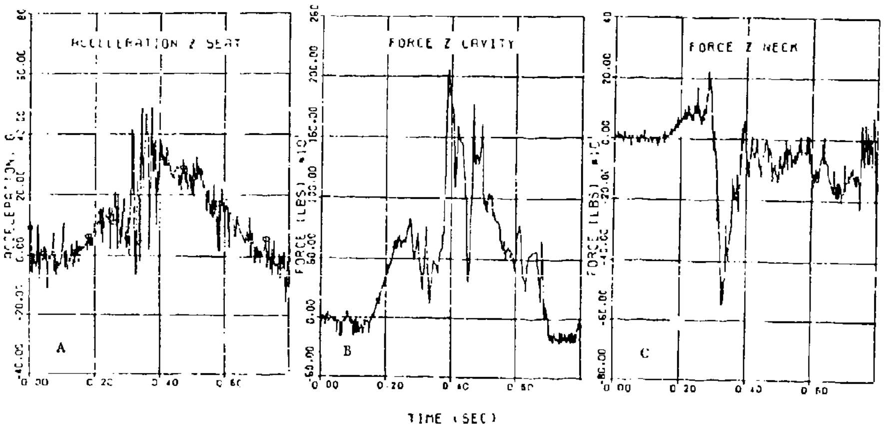

Figure 6. 720 KEAS Ejection Test. Seat and Manikin Response Parameters.

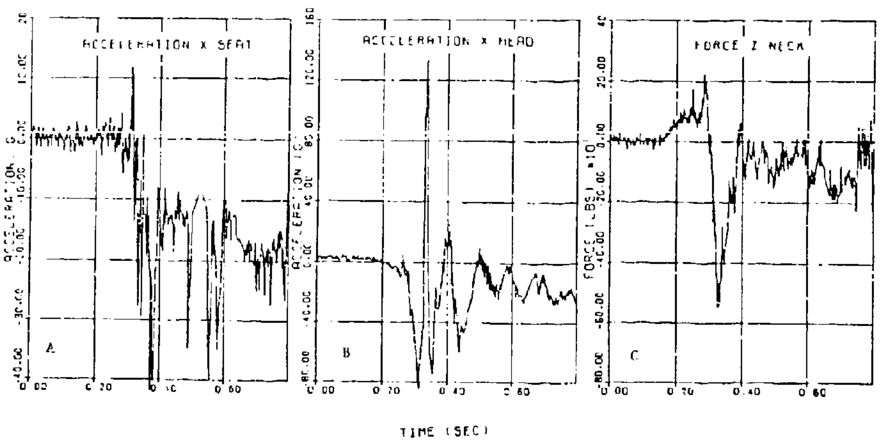

Figure 7. 720 KEAS Ejection Test. Seat and Manikin Response Parameters.

compared to live human response data. It must be pointed out that reconfigured anatomical segments shall be completely compatible with presently utilized BFM1 and consequently a one to one replacement can be undertaken without replacing the entire manikin systems presently in the inventory.

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DISCUSSION PERIOD 4

Dr Von Gierke, USA.

I have a question for Mr Frisch and other speakers, in connection with the whole topic of our symposium here on Neck Injury in Advanced Military Aircraft. We have not heard much about protection and protection matters. We talked a lot about limits and how we might be able to predict loads on the neck and probably give tolerance levels in the future, but several years ago I know that you, George, and me and several others, worked on neck protection systems. Inflatable collars and similar things at least for ejection, and we haven't heard anything about this. Has this died completely?

Mr Frisch, USA.

Quite the contrary, I thought one of the interesting things about the data that we gathered on both the low altitude as well as the supersonic ejections, was the head and neck response. Much of the tension exhibited has in the past been attributed primarily to aerodynamic lift. Now we have measures available to us to evaluate it. If I was to design a very aerodynamic helmet or inflatable collar it would not help you one iota. That is the whole point I think. We have been doing a lot of work at the high speed end with more sophisticated mannekins gathering a plethora of data to make sure that we understand what the response mechanisms are so that we can design the protective systems for the future. I think the next generation of aircraft are going to be faster, they are going to be flying higher and as an example a 1.2M escape is not unatural at all. As little as a year ago we were looking at G_z acceleration on the seat and linear and angular velocity in the dummy chest cavity. Seats are passed basically on whether or not the system hangs together, and you get a chute and don't plough into the desert. That is not a pass/fail criteria. I think we have come a long way since then. The problem is that many of the measures, although readily available, are not substantiated, and not in my opinion with known human response data. As an example torques and moments and compressive forces are very effective if I want to rank systems, if I want to establish a base line, but if someone wanted to pin me to a wall and say will 550 lbs tension injure I don't know the answer. I don't think anybody knows the answer. It has been shown to fracture cadaver cervical spines but what the relationship is to humans is is not known. However the fact that at 500kts we monitored 400lbs tension in the neck is very useful information because we positively survive 500kts ejections so consequently I would assume that the 400lbs is survivable. So now the question becomes is an additional 100lbs tension survivable. So I agree with you a lot of work needs to be done but we also have to get a better idea of what we are protecting against and what the response looks like. You have got to understand the response before you can design any protection equipment

Dr Von Gierke, USA.

I have a similar follow up question to the other problem we discussed on the neck under sustained acceleration. We have indications that the motions under sustained acceleration might be the most dangerous part. Is anyone thinking about doing anything about this. It does not necessarily have to be a neck protection system, it could for example be some experiments using mirrors to check 6 or using a display system like the super cockpit. I understand some Air Forces use mirrors for checking the rear quadrant. Has anyone experience with this or has anyone done quantification experiments on the centrifuge to explore the possibilities?

Mr Frisch, USA.

We do, and you get into some anomolies, for example the centrifuge at NADC can get pretty close to any of your aircraft performance parameters since it is close to a 12G sec^{-1} onset rate. Their can, however be problems. We ran a whole series of experiments that looked at tolerance and reclination angles and checking 6, and as an example, when high G is being pulled and test subjects turn their heads on a centrifuge, their vestibular system is upset and they are sick. Yet we continuously have pilots come in and say well I do this all the time and I don't have this problem. So either the subjects used in the experiment are not representative of the pilot population, or we need a lot of training. As an example in this presentation, the Air Force has I know had two classical hangman injuries with complete disarticulation of the spine, the Navy I think has had one, with no accompanying indication of the neck being hit. No contusions, no cuts, no marking on the helmets that would indicate that the helmet was arrested by parachute risers or garrotted or whatever. But when you start looking at data like this, the fact that you can get 550lbs extension which puts the neck into tension, and I know that you can subject that head and neck system in tension to an angular motion, could you in fact not get that kind of an injury. Again a better understanding of the response mechanisms is needed.

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14. Abstract

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Anecdotal evidence suggests that soft tissue neck injury may be an occupational hazard for fighter pilots, particularly those in the later generation trainer and combat aircraft. Recent advances in helmet-mounted devices increase the potential for injury. There is an acute need for guidance on the maximum mass and mass distribution characteristics of head-mounted devices. The Symposium discussed the extent of risk and its control through the design of helmet-mounted devices, protective systems and aircrew training and conditioning. [illegible]
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